A Portable Sensory Augmentation Device for Balance Rehabilitation Using Fingertip Skin Stretch Feedback

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***Abstract*—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 study, we proposed a form of cutaneous feedback which stretches the 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. To demonstrate the feasibility of the SAD, quiet standing on a force plate was evaluated while sensory deficits were simulated. Fifteen healthy young adults were asked to stand quietly under six sensory conditions: three levels of sensory deficits (normal, visual deficit, and visual + vestibular deficits) combined with and without augmented sensation provided by SAD. The results showed that augmented sensation via skin stretch feedback helped subjects correct their posture and balance, especially as the deficit level of sensory feedback increased. These findings demonstrate the potential use of skin stretch feedback in balance rehabilitation.**

***Index Terms*—Balance rehabilitation, biofeedback, portable device, postural control, sensory augmentation, skin stretch feedback.**

I. Introduction

POSTURAL control and balance are two crucial factors to humans in performing activities of daily living. Dysfunctional sensory systems such as vision, vestibular, and somatosensory impairments increase postural sway and the risk of falling, which threatens quality of life [1], [2]. Sensory augmentation and substitution in treatment for people with dysfunctional sensory systems have been intensively investigated in the last ten years [3]–[8]. Biofeedback systems translate bodily function information to sensory inputs such as vision, hearing, or somatosensation so that individuals are provided extra cues about their physiological states [9]. 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 [10]. Biofeedback has been known as an essential technique in rehabilitation for stroke survivors [11], [12] and the elderly [13]. 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 [10]. 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 [4], [14]. 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 [3], [5], [6], [8]. 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 [15]. 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 [16] especially at a fingertip pad, since a fingertip pad is more sensitive to skin stretch than vertical skin deformation [17]. Its easy perception, large contact surface, and the capability of providing 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. Another type of cutaneous cue, a light touch contact (contact force < 1 N) of a fingertip on a fixed surface, has been shown, in several studies, to be capable of reducing body sway in standing [18]–[21] and walking [22]. The light touch works as an additional tactile sensory input instead of a mechanical support [23]. Krishnamoorthy *et al.* [24] showed that light touch can be applied on different body parts other than fingertips to stabilize posture. Enders *et al.* [25] 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

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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. However, a portable postural corrective system using skin stretch feedback at a fingertip pad has not yet been developed or evaluated.

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. The idea was inspired by the concept of *light touch* as introduced in the previous paragraphs [20]. 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 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. This paper is organized as follows: Detailed system design and control are illustrated in Section II, results and a comprehensive evaluation on sway reduction using the developed sensory augmentation device among healthy young people are shown in Sections III and IV respectively.

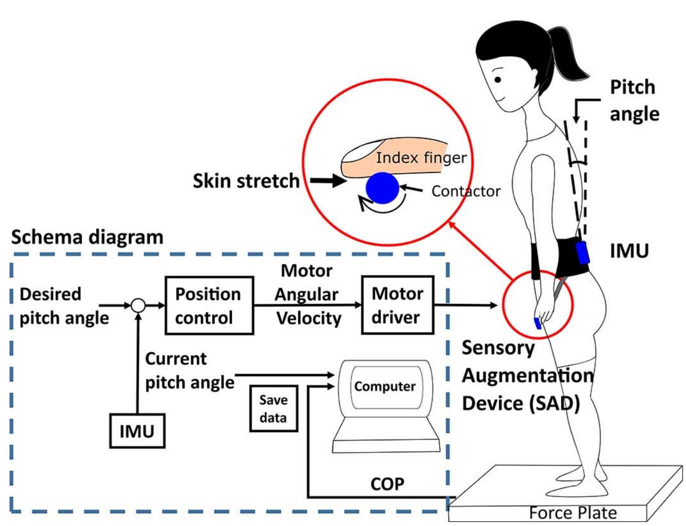

Fig. 1. Schematic diagram of sensory augmentation system. IMU measures the pitch angle of body sway while subject stands quietly on the force plate. Contactor's angular velocity is defined to be proportional to angular deviation of pitch angle from the desired pitch angle (reference angle). When subject 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 sensory augmentation device (SAD).

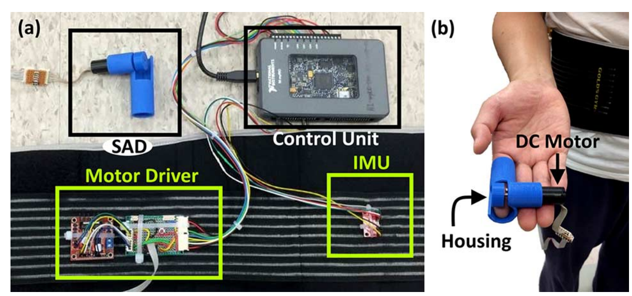

Fig. 2. (a) System consists of a sensory augmentation device (SAD) that induces the skin stretch at an index fingertip pad. A control unit, motor driver, and IMU are enclosed in a waist belt. (b) DC motor is mounted at the housing of SAD where subject's index finger is inserted. Cutaneous skin stretch feedback is therefore provided by the shearing between contactor operated by the dc motor and fingertip pad.

II. Materials and Methods

A. Device Development

The schematic diagram and the fabricated device of our sensory augmentation system are illustrated in Figs. 1 and 2, respectively. The device's detailed design and related control strategy are described in the following subsections.

1) Design of Portable Sensory Augmentation Device (SAD): SAD was designed to induce skin stretch at an index fingertip pad [Fig. 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. 2(b)]. Skin stretch feedback was therefore provided by the shearing between the contactor, operated by the dc motor, and the fingertip pad [Fig. 2(b)]. Several contactors and housings of various sizes were fabricated to accommodate various subjects' finger sizes; we created these using a 3-D printer (Replicator 2X, Makerbot, Brooklyn, NY, USA). 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 inertia measurement unit (IMU) (MPU-9150, InvenSense Inc., San Jose, CA, USA) 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) [Figs. 1, 2(a)]. The data from the IMU were then used to monitor the postural sway of the subject during quiet standing. An algorithm developed by Madgwick [29] 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, USA) 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 (Figs. 1–3). We used an h-bridge type motor driver (L298N, STMicroelectronics, Italy) to provide the appropriate amount of power for the dc motor [Fig. 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 a stable estimate of COM displacement. Sampling rates of SAD, and IMU were 1 kHz, and 500 Hz, respectively [Figs. 1, 2(a)].

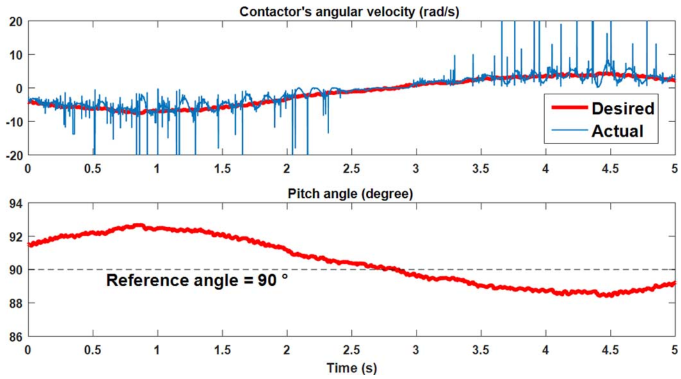

Fig. 3. Relationship between contactor's angular velocity and pitch angle of subject. In this example plot, the reference angle was set to 90°.

2) Control Strategy: 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. Figs. 1 and 3 show 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 presence in actual velocity (Fig. 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.

B. Experimental Protocol

Fifteen healthy young adults (four females and 11 males; mean age $\pm$ s. d.: 26.4 $\pm$ 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 SAD. This study was approved by the Texas A&M University Institutional Review Board.

Subjects were asked to stand quietly on a force plate (OR6, AMTI, Watertown, MA, USA) 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 [30]–[33]. 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 [30], [31]. 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. 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 Figs. 1 and 2(a)].

The experiment consisted of two parts: practice session and 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 ten 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.

C. Postural Sway Measures

A force plate (OR6, AMTI, Watertown, MA, USA) and a data acquisition system (DAQ) (USB-6002, National Instruments, Austin, TX, USA) 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 [34]. Many studies have evaluated the postural steadiness based on a single measurement [2], [23], [35]. However, it may not be sufficient since some postural sway measures are not sensitive enough to distinguish various aspects of postural impairment [36]. In this study, multiple traditional COP measures were investigated both in time domain and frequency domain [34]. 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 a 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 [34]. 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.

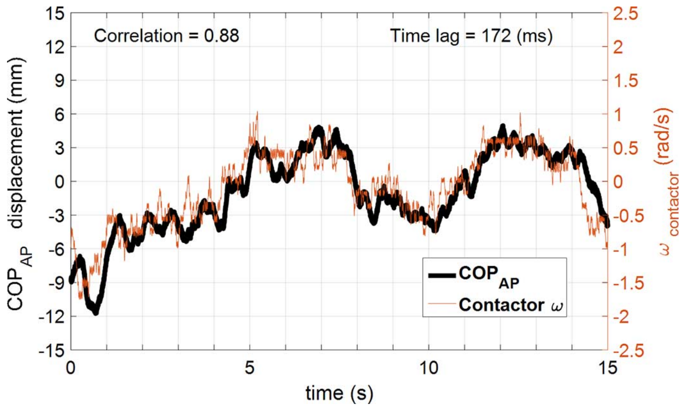

Fig. 4. Time series of COP displacement in AP direction (black bold line) and contactor's angular velocity over 15-s period. The data was obtained from the same subject (subject 6) in VVD condition. Positive correlations ($r = 0.88$) and positive time lag (172 ms) are shown indicating that skin stretch ($\omega_{\text{contactor}}$) is ahead of COP$_{\text{AP}}$ 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.

D. 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 $\alpha = 0.05$ (SPSS, v21, Chicago, IL, USA). The cross-correlation (XCORR) function was used to identify the time delay between contactor's angular velocity and COP$_{\text{AP}}$ time series. Correlation coefficient between two time series was also calculated using MATLAB (R2015a, MathWorks, Natick, MA, USA).

III. Results

Fig. 4 shows the correlation between skin stretch (SAD) and COP in AP direction. Fig. 5 shows four postural sway measures of COP data in both AP and ML directions across 15 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 I, 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.

A. Correlation Between Skin Stretch and COP_{AP}

The time series of COP displacement in AP direction and angular velocity of a contactor, denoted by $\omega_{\text{contactor}}$, is depicted in Fig. 4. The example data was selected from one of the subjects (subject 6) in VVD condition. Skin stretch on the fingertip pad was generated by the contactor as it rotated at $\omega_{\text{contactor}}$. Hence the level of skin stretch can be represented by $\omega_{\text{contactor}}$. Since the skin stretch was applied only based on AP direction, we examined COP displacement in AP direction only. The result shows that COP$_{\text{AP}}$ movement correlates $\omega_{\text{contactor}}$ with $r = 0.88$ and time lags of 172 milliseconds. The average correlation r and time lag of 15 subjects are 0.82 (s. d. $= 0.15$) and 150 ms (s. d. $= 22$ ms), respectively. It indicates that skin stretch ($\omega_{\text{contactor}}$) is ahead of COP$_{\text{AP}}$ movement suggesting that SAD led the postural sway of the subject in quiet standing.

B. Effect of Sensory Deficits

All parameters except MF_{ML} ($p > 0.05$) indicated significant differences among three sensory modality conditions as shown in Table I. Range$_{\text{AP}}$ 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$_{\text{AP}}$: $p < 0.001$; MV_{AP}: $p < 0.001$). Range$_{\text{ML}}$ of postural sway was greater in VVD compared to ND (Range$_{\text{ML}}$: $p = 0.007$). 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}: $p = 0.009$).

C. Effect of Sensory Augmentation

From Table I, 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}: $p = 0.002$),

D. Interaction Effects of Sensory Deficits $\times$ Sensory Augmentation

The analysis revealed a significant interaction effect between sensory modality and sensory augmentation in Range$_{\text{AP}}$ ($p = 0.019$) as presented in the right-most column in Table I. While applying skin stretch feedback, Range$_{\text{AP}}$ tended to decrease in VVD, whereas it tended to increase Range$_{\text{AP}}$ in ND and VD when compared to when SAD was turned OFF. Pairwise comparisons revealed that SAD significantly increased Range$_{\text{AP}}$ ($p = 0.037$) in ND. Similarly, SAD provided a 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$_{\text{ML}}$, MF_{AP}, CF_{AP} and CF_{ML}.

IV. Discussion

The main contribution of this study is that the developed sensory augmentation system was able to detect the body sway angle by the integrated IMU and provide an additional cue of postural sway by SAD. Unlike the existing techniques that

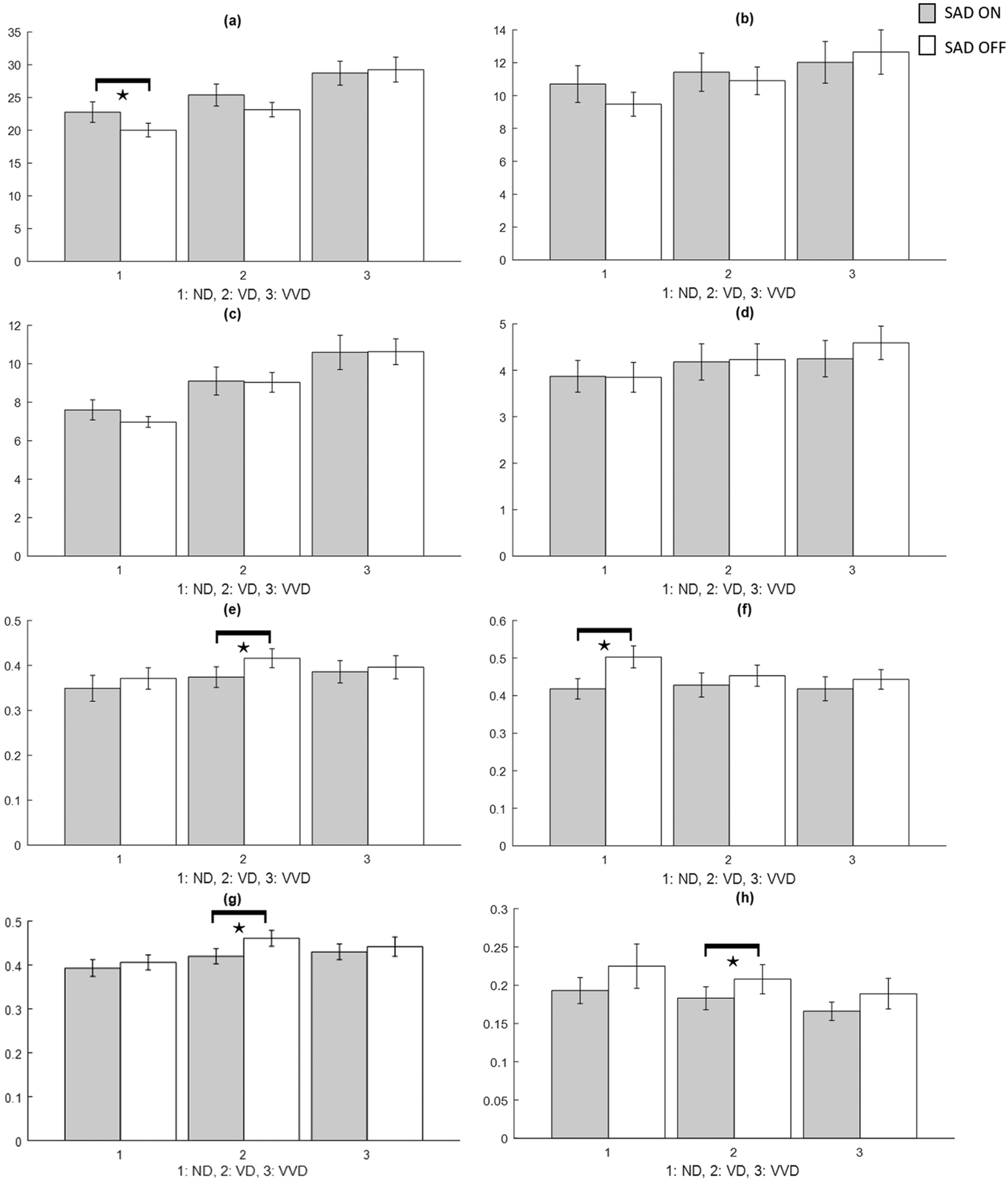

Fig. 5. Mean values for range, mean velocity, mean frequency, and centroid frequency of COP in both AP and ML directions for each of ten trials under three sensory deficit conditions (1: ND, 2: VD, 3: VVD). Each condition shows when SAD is turned ON (Grey) and SAD is turned OFF (White). Error bars indicate 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. (a) Range of CoP in AP (mm). (b) Range of CoP in ML (mm). (c) Mean Velocity of CoP in AP (mm/s). (d) Mean Velocity of CoP in ML (mm/s). (e) Mean Frequency of CoP in AP (Hz). (f) Mean Frequency of CoP in ML (Hz). (g) Centroid Frequency of CoP in AP (Hz). (h) Centroid Frequency of CoP in ML (Hz).

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

TABLE I
MEASURES OF POSTURAL SWAY. VALUE REPRESENTS MEAN (STANDARD DEVIATION) FOR THREE SENSORY MODALITY CONDITIONS AND TWO SENSORY AUGMENTATION CONDITIONS, AND INTERACTION (Sensory Modality × Sensory Augmentation) *p*-VALUES

Parameters	Sensory Modality Conditions			Sensory Augmentation		Interaction *p*-value
	No Deficit (A)	Visual Deficit (B)	Visual & Vestibular Deficit (C)	SAD ON (D)	SAD OFF (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

Superscript denotes significant differences from indicated main effect condition ($p < .05$).

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 effect of induced skin stretch feedback at the fingertip may seem to be contradictory. 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. 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.

First of all, 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}. 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. 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} ($p = 0.027$).

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. 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 [37], [38] . 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 [39]. *CF* may give us an insight on how well the postural control could be achieved under different task constraints [35]. 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 [38], therefore enhancing quiet standing postural control.

Removing sensory information (VD) or challenging balance condition (VVD) significantly increased postural sway, which agrees with the previous studies [40]–[44]. As expected, when all the sensory systems are functional, individuals' postural control was significantly better, compared to when there were not 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.* [34] 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 happen 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.

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 [45], [46]. 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.

In the literature, *MV* was suggested as the most significant measure for separating different groups (e.g., age) [34] and the most reliable among traditional parameters [47]. 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.

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 [25], [48], [49]. Manjarrez *et al.* [50] 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 [25], [48], [49]. Jeka *et al.* found that COP displacement [18], [20] and left leg EMG activity [19] 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 [51], [52]. Nashner [53] found that a long-latency postural reflex (120 ms) helps to reduce postural sway, which is usually classified as a supraspinal pathway [54], [55]. In our study, the time lag was approximately 150 ± 22 ms (mean value $\pm$ 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 [56]. 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 [18], [20], [21], [24], [55] required reachable fixed surfaces or sizeable 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.

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. Different populations (e.g., the elderly or patients with balance disorders) can be examined instead of healthy young adults with simulated sensory deficits Furthermore, a future study will investigate the effect of various locations (e.g., wrist) of skin stretch feedback on balance. This is because applying skin stretch feedback at the fingertip may hinder the use of the hand and fingers and eventually activities of daily living.

V. Conclusion

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.

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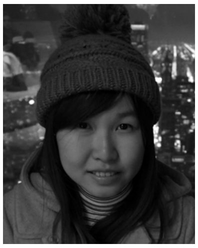

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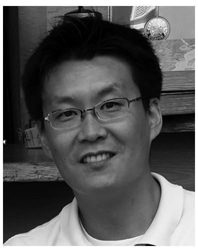

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