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Sensory garments with vibrotactile feedback for monitoring and informing seated posture

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Abstract— **Increases in the number of sedentary workers and the rate of related musculoskeletal injuries have spurred interest in posture monitoring and feedback devices. The emergence of flexible sensors that can be integrated into wearable garments provides a unique solution to unobtrusive monitoring of postural movements. Furthermore, a postural guidance system can be realized with the addition of a feedback display. In this work, a flexible strain sensor was integrated with a Raspberry Pi processor and a vibrotactile feedback display to form a sensory feedback garment that measured, evaluated, and communicated real-time feedback of seated lumbar posture. The sensor was evaluated for measurement accuracy via comparison to a Vicon motion capture system, and the garment was evaluated for efficacy of feedback guidance. Across studies of five participants, we found that the sensory garment reproduced the lumbar angle with low error (5.4%) relative to a motion capture system, and that providing haptic feedback resulted in participants maintaining lumbar posture within customized target zones for substantially more time than in the absence of haptic cues.**

I. Introduction

As of 2003, musculoskeletal complaints were the second most common cause of short term and most common cause of long term medical absences from the workplace (10-20% of primary care consultations) and were the most expensive disease category in a cost of illness study [1]. Computer workstation posture, especially when static, has been shown to be a significant contributor to the development of musculoskeletal ailments [2]. While there is no agreed upon proper posture during sitting, there are many postures that are deemed to be poor and should be avoided [3]. Seated postural training programs have proven effective, however they require participants to self-monitor and lead to decreases in productivity [4]. Furthermore, posture monitoring systems used in clinics or labs require large or expensive equipment and are not suitable for everyday use [5].

Typical biofeedback systems employed for postural or balance guidance and correction employ a sensor (or network of sensors) to capture body movement, and a display to relay motion tracking information and corrective cues. Recently, several portable or wearable systems have been developed to monitor posture and, in some cases, provide corrective feedback. Zheng, *et al.* employed force sensors embedded in office chairs to detect seated posture and used visual and vibrotactile cues to encourage users to adopt specific reference postures [6], [7]. Dunne, *et al.* employed an optical fiber sensor to monitor seated posture while performing computer workstation tasks and compared the results to clinical opinions of posture, though did not provide corrective feedback [8]. Wang, *et al.* used two inertial measurement units to monitor thoracic spinal angles and provided visual feedback during torso flexion-extension exercises via a smartphone app [9]. Wong, *et al.* used a network of three inertial measurement units to measure thoracic and lumbar angles and provide auditory feedback to encourage a neutral standing position [10].

Additionally, wearable posture trainers are also gaining traction in commercial markets, with products such as the UpRight Go Posture Trainer, Lumo Lift, Prana, Alex, and Mevics posture monitors currently available. Most provide feedback via vibrotactile cues, allow for historical posture tracking through app-based visual graphics, and are attached to the skin or clothing via adhesive, clips, magnets, or pins. These devices generally rely on inertial measurement units to determine posture from a single node, thereby limiting postural reconstruction to a single vector relative to gravity and are not capable of directly measuring relative joint angles.

The emergence of soft, stretchable sensors offers a unique solution space for wearable feedback devices, whereby garments with integrated sensing and actuation can potentially be made non-obstructive or imperceptible to the user. Soft sensors are typically composed of low-modulus materials that transduce physical deformations to electrical signals. Various types of these sensors developed for detecting human motion include silicone elastomers with embedded liquid metal microchannels [11], [12], conductive polymer composites for resistive [13] and capacitive sensing [14], stretchable optical fibers [15], and conductive textiles [16], [17].

Systems utilizing flexible strain sensors (FSS) have previously been successfully employed to reconstruct upper body poses and detect kinematic angles. Mattman, *et al.* developed a posture monitoring system using a network of FSSs in conjunction with a classification algorithm to detect exercise movements and 27 discrete upper body poses, reporting 97% classification accuracy during pose recognition [16], [17]. Yamamoto, *et al.* investigated the use of FSSs adhered directly to the back for tracking tri-axial movements of the lumbar spine, reporting errors of less than 3° in estimated lumbar angles for flexion-extension, side bending, and rotational movements [18]. Fujimori, *et al.* developed a wearable motion capture suit that integrated flexible tactile

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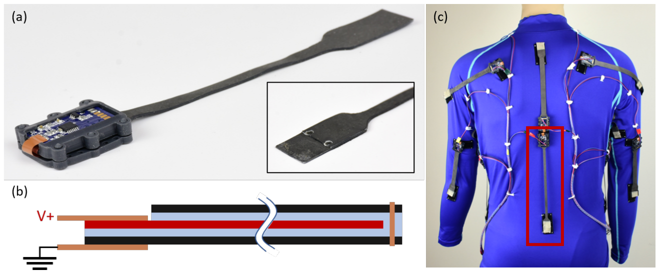

Fig. 1. (a) A photo of the capacitive sensor. The inset shows the end of the sensor where it is interfaced with the signal conditioning electronics board via copper strips. (b) Schematic of the cross-section of the sensors. The red layer indicates the active electrode, the black layers indicate the ground electrodes. These conductive layers are separated by dielectric material, indicated in blue. Copper wire is used to electrically connect the upper and lower ground electrodes, and copper strips are affixed to the surface of the active and one of the ground electrodes to interface with the signal conditioning boards. (c) Photo of the sensors integrated into an upper-body sensory garment. This study focused on the efficacy of the sensor situated on the lower back, highlighted in the red box.

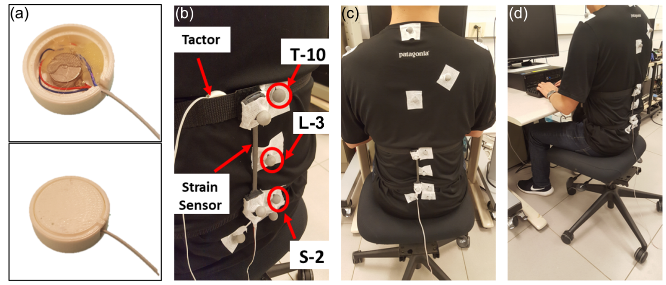

Fig. 2. (a) Tactor assembly showing coin-style vibration motor mounted inside plastic housing (top) and fully assembled (bottom). (b) Flexible strain sensor, reflective markers, and tactor mounted on a participant. (c) Rear view of participant seated in backless chair at computer workstation. (d) Seated position of participant at computer workstation.

sensors with an inertial measurement unit that was able to classify several poses [19]. Though these findings are limited, it is evident that soft sensors have the potential to monitor postural pose and determine angular relations between segments of the spine.

To provide cues to the user, feedback displays can be designed for visual, auditory, or tactile interactions. However, it is imperative to consider the impact of display modality on intended use and ease of interaction with the system. Unlike visual and auditory displays, providing feedback via vibrotactile cues avoids interference with vision and hearing. Vibrotactile displays have been implemented to prevent collisions and provide navigation information while driving [20]–[22], to provide altitude information, warning signals, and replace or reinforce visual and auditory cues while flying [23], [24], and to enhance virtual reality environments [25], [26]. For biofeedback purposes, vibrotactile displays allow for fast encoding of directional cues and can be co-located with measurement devices to provide more intuitive operation [27], [28]. Furthermore, vibrotactile displays have been successfully implemented in arm motion training, balance training, and gait rehabilitation programs to provide corrective cues in real time [29]–[35]. Therefore, these types of displays are well-suited for integration into a wearable biofeedback system for indicating postural deviation.

II. Materials and Methods

A. Sensory Feedback Garment

The sensory feedback garment consisted of a flexible strain sensor (FSS), custom PCB, Raspberry Pi processor, and tactor. The FSSs employed in this work are an extension of the conductive composite-based capacitive sensors presented originally by White, *et al.*, in which the sensors were demonstrated to be reliable for thousands of cycles and withstand strains up to 275% with no change in functionality [36]. In this work, the previous three-layer parallel plate capacitor structure of White, *et al.* was modified with the addition of another ground and dielectric layer, resulting in a five-layer capacitor, as discussed in [37] (Figure 1). The additional layers result in the active layer being fully surrounded by ground planes, which increases the signal to noise ratio by reducing the influence of electrostatic noise on the sensor output via shielding and by multiplying the capacitance of the sensor by a factor of four relative to a three-layer sensor of the same size. The center electrode layer was composed of a silicone elastomer (DragonSkin 10 Slow, Smooth-On, Inc.) blended with expanded graphite (Expandable graphite, Sigma Aldrich), and separated two dielectric layers composed of the unmodified silicone elastomer. The ground planes comprising the top and bottom outermost layers of the sensor were made of the same conductive composite used for the center electrode.

The conductive composite material was made by mixing silicone with expanded graphite particles suspended in cyclohexane (a solvent for silicone) for a final loading of 10wt% graphite in the cured silicone, after the cyclohexane fully vaporizes (more details are discussed in [36]). The ground electrode, dielectric, and active electrode layers were rod-coated using a threaded rod to create large films ($\approx$ 25 cm $\times$ 40 cm), building up the capacitive structure layer-by-layer. After rod-coating the active electrode, the film was folded onto itself to create the 5-layer structure. The sensors were then laser cut from the film, cleaned with soap and water, and then interfaced with the signal conditioning electronics. A custom PCB attached to the end of the sensor measured the change in capacitance between the electrode layers as strain was applied (Figure 1a). Output from the PCB was transmitted to a microcontroller (Raspberry Pi 3 Model B) for inline data processing using custom software developed in Python. Sensor signals were collected at 25 Hz and low-pass filtered at 2 Hz to remove high frequency noise not associated with body motion.

A tactor, consisting of a coin-style vibration motor seated in a custom plastic housing (Precision Microdrives, 310-101 vibration motor encased in plastic housing [38]), was also connected to the microcontroller (Figure 2a). The vibration motor had a frequency of 200 Hz at an operating voltage of 3.0 V and had a spin-up time of 90 ms. The motor was installed in a 3D printed housing measuring 27 mm in diameter by 8 mm thick which served to increase skin contact and amplify the feedback sensation. The tactor was set to vibrate when certain strain levels were exceeded by the FSS. Strains thresholds were customized for each participant based on lumbar range of motion. Tactor activation was logged by the FSS system.

B. Motion capture setup

A Vicon motion capture system (Giganet MX with T-20 Series cameras (10), Vicon) was used to measure reflective marker positions for calculation of lumbar spine angles. FSS and Vicon data collection was synced electronically at the start of each trial. Vicon motion capture data was collected at 50 Hz. 3-D marker positions were extracted using Nexus 1.7.8 (Vicon) and imported to MATLAB for post processing. All angular measures were calculated during post-processing.

C. Subjects

Five young (26.2 $\pm$ 1.3 yrs) healthy individuals (3 female, 2 male) recruited from the University of Michigan student body participated in this study. The University of Michigan Institutional Review Board approved the experimental protocol, and researchers obtained written informed consent from each participant prior to the start of the experiment.

D. User study test procedure

All participants completed a single testing session. After providing informed consent, participants donned a polyester exercise shirt. A FSS was secured to the lumbar region of the back using Velcro straps. The ends of the sensors were fastened dorsal to the T-10 and S-2 vertebrae. Three reflective markers were affixed alongside the sensor, such that one marker corresponded to the transverse body plane at the T-10, L-3, and S-2 vertebrae. The markers were positioned as close to the spine as possible with cloth tape. A single tactor motor was placed at the T-10 end of the sensor and was held in place with Velcro strapping. An electrode pad was affixed to the left elbow and connected to the ground pin of the microcontroller. The experimental setup is shown in Figure 2b-d.

Participants sat in front of a computer workstation in a backless office chair. The height of the chair was adjusted such that the thigh and shank were approximately perpendicular when seated. The height of the computer monitor was adjusted such that participant line-of-sight was parallel to or angled slightly below horizontal. Desk height was fixed.

At the start of each trial, participants were asked to sit tall with what they believed to be proper posture, and to slouch the lower back, not the shoulders. Measurements were taken in both positions and were used to characterize the lumbar range of motion. A feedback threshold was then set to 90% of the range of motion as measured by the FSS (where 100% represented the participants proper posture). If a participant slouched to a point below the 90% threshold, the tactor was activated to provide an out-of-posture cue until posture was adjusted back to the 90%-100% range. Tactor cues only provided alerts that the participant had deviated from the desired position; they did not indicate directional errors. Participants were instructed to make corrections until vibrations from the tactor ceased. Participants were not allowed to practice with feedback prior to testing but were instructed to sit tall and attempt to return to their own proper posture to stop a tactor cue. During testing, participants were tasked with retyping a document displayed on a portion of the workstation screen. Instructions were to transcribe as quickly and accurately as possible, while responding to any tactor cues with a posture correction. Participants were instructed to look only at the monitor, and to refrain from leaning on the desk.

Participants completed two trials, each lasting 10 minutes. During the first trial, tactor activation was disabled for the first five minutes (Block 1, feedback (FB) Off) and enabled during the second five minutes (Block 2, FB On). During the second trial, tactor activation was enabled during the first five minutes (Block 3, FB On) and disabled for the second five minutes (Block 4, FB Off). Participants were only asked to respond to cues and were not aware of when tactors were enabled or disabled. There was an approximately two-minute break between trials, and the text to be transcribed was changed for the second trial.

Following completion of testing, participants responded to eleven prompts on a comparative Likert scale survey (strongly disagree (1), disagree (2), neutral (3), agree (4), or strongly agree (5)) covering the use and instructions for use of the posture feedback system. When responding to prompts, participants were instructed to focus on their interactions with the FSS and tactor.

E. Data Analysis Methodologies

Three-dimensional positions of the T-10, L-3, and S-2 markers were used to calculate the lumbar angle as measured by the Vicon cameras. The motion capture data was then resampled at the FSS time points using the interp1 function in MATLAB to account for collection rate differences. A linear regression was fit to determine the root-mean-square error for each trial between the FSS output and the calculated lumbar angle. Percent time in the target posture zone (PIZ, 90-100% of ROM) was calculated for each five-minute block within a trial (FB enabled, FB disabled). Paired t-tests were used to evaluate differences in PIZ for the feedback on and feedback off conditions, and to evaluate learning effects between the first feedback off and the last feedback off condition. Responses to the survey were evaluated on a five-point scale and averaged across participants to gauge sentiments towards device usability and identify any potential bias introduced by instructional deficiencies.

III. RESULTS

A. Flexible Strain Sensor Performance

Representative linear regressions between FSS and lumbar angles for the two trials of all participants can be seen in Figure 3, while time series representations for the same trials,

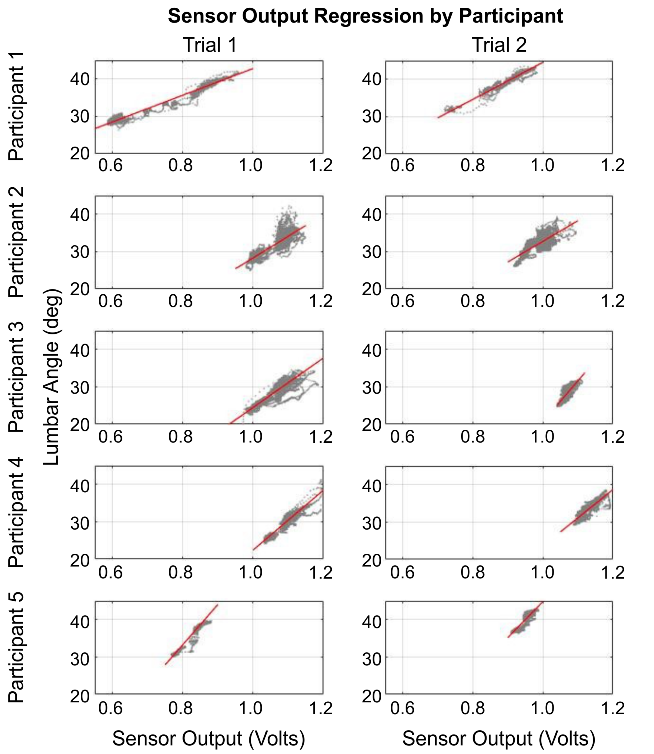

Fig. 3. *In situ* sensor calibration with respect to lumbar angle. Linear regression plots showing comparison of flexible sensor output versus the lumbar angle as measured by the Vicon motion capture system. Regressions for all trials and participants are shown.

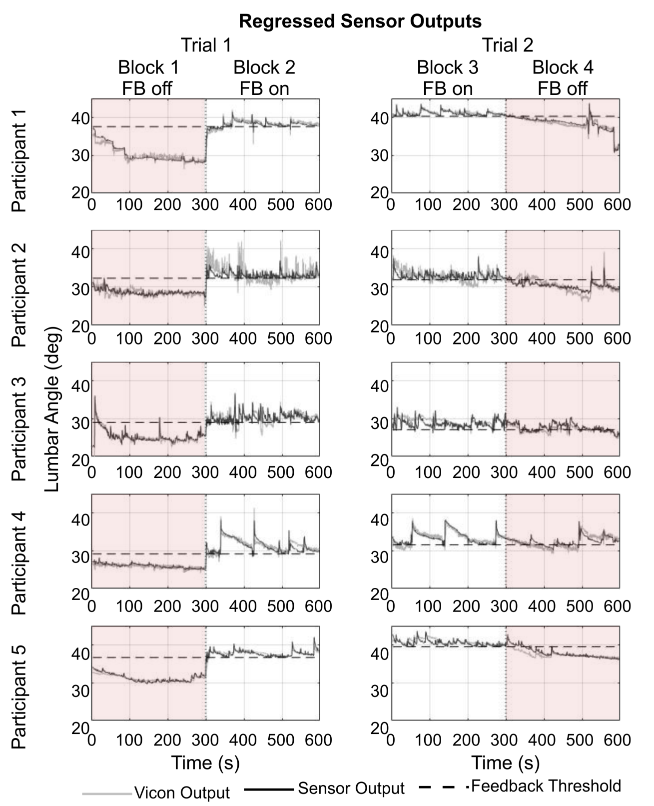

Fig. 4. Flexible strain sensor and Vicon lumbar measurement outputs for all trials. Sensor outputs are displayed as lumbar angle in degrees after transformation via linear regression. Feedback thresholds and block dividers are shown via dashed horizontal and dotted vertical lines, respectively.

#	Statement	Score
P1	Satisfied with the sensory garment	4.8
P2	Satisfied with the ease of use of the garment	4.6
P3	Able to complete tasks quickly with the garment	4.4
P4	Felt comfortable using the garment	4.4
P5	Easy to learn to use the garment	4.8
P6	The information conveyed by the tactor was clear	4.6
P7	Could recover easily when a mistake was made	4.6
P8	Could recover quickly when a mistake was made	4.6
P9	The instruction provided by the tactor was clear	4.8
P10	The instruction provided by the tactor was effective in completing the given task of maintaining a certain lumbar angle	4.2
P11	The instruction provided with the garment was easy to understand	4.6

TABLE I

EVALUATION OF USER SATISFACTION WITH THE GARMENT. THE MEAN SCORE OUT OF 5 CORRESPONDING TO THE LIKERT SCALE IS LISTED.

with sensor outputs converted from voltage lumbar angle via the regressions, are shown in Figure 4.

The average lumbar angle as measured by the motion capture system was 32.8° ± 4.8°, while the average trial minimum lumbar angle was 28.7° ± 5.2° and the average maximum was 37.3° ± 4.9°. The average range of motion minimum was 19.5° ± 5.9° and the maximum was 35.0° ± 4.8°. The average feedback threshold, below which the tactor would provide a cue, was 33.5° ± 4.7°. The mean FSS root mean square (RMS) deviation from motion capture measurements across all participants and trials was 0.85° ± 0.26°, and the average R^2 among regressions was 0.83 ± 0.14. The average regression slope was 73.4 ± 25.1 degrees per volt. There was significant variability observed in the regression slopes between trials ranging from 35.7 to 109.4 degrees per volt. The largest within-participant difference between trials was 43.2 degrees per volt, though all other participants had slope differences of less than 14 degrees per volt. We hypothesize that this is due to shifting of the sensor on the body and changes in the integrity of the interface between the sensor body and the signal electronics.

B. Postural Feedback Efficacy

The average percent time in the target zone (PIZ) for each five-minute testing block was 0.9 ± 1.9%, 94.4 ± 6.3%, 99.1 ± 0.4%, and 35.5 ± 34.8% (Block 1 FB Off, Block 2 FB On, Block 3 FB On, Block 4 FB Off, respectively; see Figure 5). Paired t-tests indicated no significant differences in PIZ between feedback enabled blocks (Blocks 1 and 4, $p>0.05$) or between feedback disabled blocks (Blocks 2 and 3, $p>0.05$). There were significant differences in PIZ between feedback on and feedback off blocks within a trial (Blocks 1 and 2, $p<0.0001$; Blocks 3 and 4, $p<0.0001$).

C. Subjective Evaluation of Sensory Feedback Garment

Results of the Likert survey are shown in Table I and indicate an overall positive reaction to the sensory feedback garment. Average responses indicated agree to strongly agree sentiments for all prompts. We concluded therefore that the stiffness of the FSSs was sufficiently low and the size of the

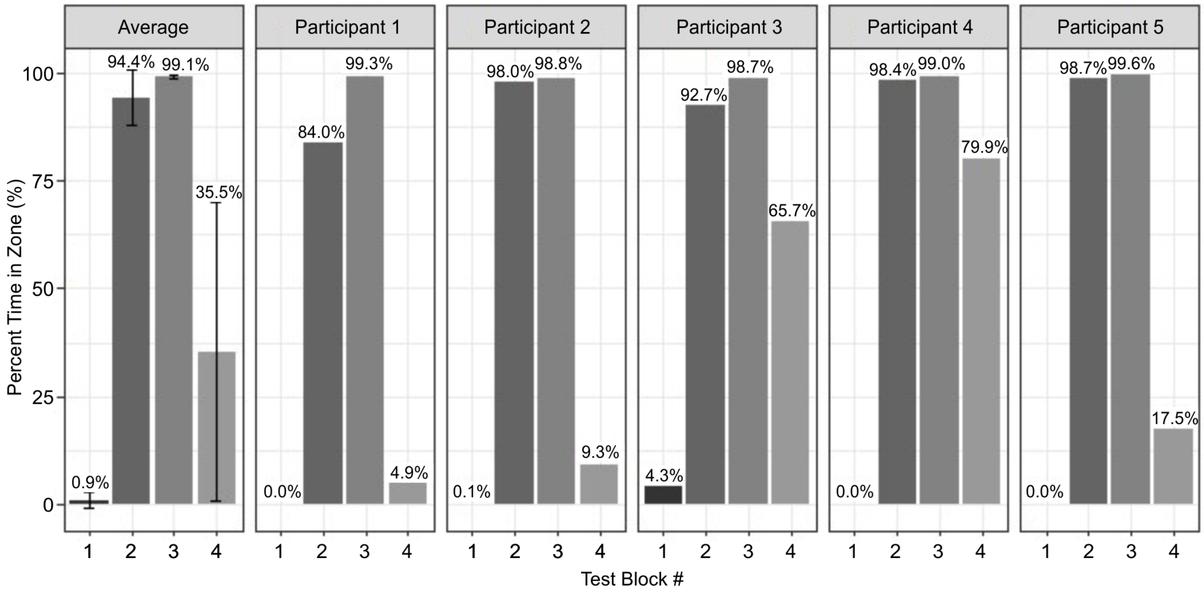

Fig. 5. Percent time in the postural target zone (PIZ) in the feedback on (blocks 2 and 3) and feedback off (blocks 1 and 4) conditions. The results from each particplant are shown; the mean PIZ values are shown in the leftmost chart. A marked improvement in PIZ is observed when vibrotactile feedback is provided.

tactor small enough so that the garment was perceived to be comfortable and sufficiently imperceptible to the wearer of the garment. The results of this subjective evaluation thus indicate that materials and form factor of the sensor garment are a suitable basis to build upon from a comfort and compatibility perspective, and point to areas in which the garment can be improved.

IV. Discussion

This study aimed to characterize the efficacy of a flexible capacitive strain sensor for measurement of seated lumbar angle, and to evaluate the utility of a postural feedback system during seated workstation tasks. A motion capture system was used as a measure of true lumbar excursion to validate FSS measurements. Lumbar angles calculated from motion capture data agreed with those measured in previous work, indicating that seated postures during the typing tasks were within normal ranges [39].

Linear regressions within trials yielded a mean RMS error of 0.85° over a 15.5° range of motion, or a 5.4% average error. In comparison, Yamamoto, *et al.* reported RMS errors of 2.35° for flexion/extension of the lumbar spine using a network of flexible sensors [18], and Wong, *et al.* reported RMS errors of less than 3.1° using a system comprising several inertial measurements units [10]. In the case of Yamamoto, *et al.*, flexion angle was determined as the average output of two flexible sensors located 10 cm laterally from the spine whereas the single sensor used in this study was positioned directly dorsal to the spine, which may have contributed to the smaller observed RMS error. Wong, *et al.* employed inertial measurement units which were subject to drift errors that were algorithmically corrected, while the strain sensors in this study were less subject to drift. With regard to efficacy, 0.84° is well within the 2-5° error that McGinley, *et al.* and Cuesta-Varagas, *et al.* recommend for clinical interpretation [40], [41]. However, minimum R^2 values for linear regressions were smaller than other researchers have reported, with the minimum R^2 for a single trial being 0.57. Only five trials (out of 10) had R^2 values above 0.85, and regression slopes were variable across participants and trials. Variability of regression slopes may be attributable to differences in system capacitance, which is slightly different for each user based on sensor-to-body contact and the differences in electrical activity of the body.

Percent time in target zone results indicate that participants were able to successfully use vibrotactile feedback about their lumbar angle to prevent flattening or rounding of the lower back while completing a typing task. During Block 1 of the first trial, when feedback was disabled, participants spent nearly no time in the target zone as they slouched almost immediately following calibration. When feedback was activated in Block 2, participants were able to respond quickly and spent most of the block in the target zone. Blocks 3 and 4 (Trial 2) were included to identify if any learning effect was present after the short training period, and any learning effect would be evidenced by an increase in PIZ between Blocks 1 and 4. During Block 3 participants again remained in the target zone for over 99% of the block, with time in zone dropping off after feedback was again disabled in Block 4. While there was no statistically significant difference between Blocks 1 and 4, there was a trend towards and increased residence in the target zone, with the average PIZ increasing from 0.9% to 35.5%. Furthermore, all participants experienced increased residence time in the target zone from Block 1 to Block 4. This trend indicates that implementation of a training regimen may improve overall posture over time. Research in motor learning has shown that the amount and timing of feedback presented are important to consider when developing a training program, and these principles should be considered when designing postural feedback training programs [42]–[44].

The survey results indicate that most participants felt that the device was easy to use and communicated deviations from the target lumbar angles well, even with a single tactor feedback display (Table I). Furthermore, the simple instruction provided prior to use of the system was sufficient to elicit an understanding of how the device functioned and proper responses to vibrotactile cues, even without the implementation of a practice session. Anecdotally, verbal comments made to the research team indicated that participants did not notice the FSS but did find the Velcro strapping used to position the sensor to be a source of some discomfort. The use of Velcro fasteners was employed to allow for adjustability of the system to ensure proper placement of the sensors relative to the spine and specific vertebral levels.

Overall, the results of this study indicate that a single flexible strain sensor is an effective tool for monitoring seated lumbar angle and can be used in tandem with vibrotactile feedback as both an aid and training tool to guide seated posture. To our knowledge, this is the first study to employ an elongation sensor with vibrotactile cuing capabilities to monitor and guide seated posture. Further development of this technology has enormous implications for workplace health and safety, as soft sensors of this nature can be incorporated directly into clothing, allowing eventual widespread monitoring of daily posture that could elucidate trends in the development and treatment of musculoskeletal disorders common to sedentary workplace settings. This would allow for evidence-based development of postural training programs that employ vibrotactile feedback to encourage beneficial postural habits.

This study was not without limitations. The small population size and age demographic are useful for proving the efficacy in a pilot study but limits the statistical power of the results. The participant population must be expanded to better understand sensor and feedback system behavior and reception across the age and body-type spectrum. The sensor and feedback system were also tested in an ideal setting whereby motion was mainly limited to torso flexion and extension in the sagittal plane. Therefore, the efficacy of the sensor is unknown in more commonly experienced situations where there is also flexion in the coronal plane and torsion along the spinal axis. Future studies should then focus on improvement to measurement capabilities beyond a single direction of motion and may require the development of a network of sensors to improve posture reconstruction accuracy. Additionally, future work should aim to characterize the effect of varying sensor placement on measurement errors, as sensor location is likely to vary as sensors are embedded directly into garments such as shirts.

V. Conclusion

This work aimed to develop a standalone system for monitoring posture and providing postural guidance, and subsequently evaluated the efficacy of the system for both postural measurement and postural guidance efficacy. A flexible strain sensor was paired with a vibrotactile display and implemented to measure lumbar angle and provide corresponding postural feedback. The results indicate reasonable measurement errors when compared with other posture monitoring devices, and current acceptable errors observed in clinical settings. Furthermore, participants successfully utilized postural feedback to maintain lumbar angles in target zones. These findings indicate that the device effectively provides postural aid and shows promise as a postural trainer. Further research should focus on implementation of multi-sensor networks to potentially improve measurement quality and the impact of training with postural feedback devices.

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