Effects of Vibrotactile Biofeedback Coding Schemes on Gait Symmetry Training of Individuals With Stroke

Muhammad Raheel Afzal, *Member, IEEE*, Hosu Lee, Amre Eizad, Chang Han Lee,
Min-Kyun Oh, and Jungwon Yoon, *Member, IEEE*

***Abstract*—Variations in biofeedback coding schemes for postural control, in recent research, have shown significant differences in performance outcomes due to variations in coding schemes. However, the application of vibrotactile biofeedback coding schemes to gait symmetry training is not well explored. In this paper, we devised various vibrotactile biofeedback modes and identified their efficacy during gait symmetry training of individuals suffering from hemiparesis due to stroke. These modes are composed of variations in vibration type (on-time or intensity), and relation type (proportional or inversely-proportional) with the error in symmetry ratio. Eight individuals with stroke participated in walking trials. From dependent t-tests on the collected data, we found improved achievement of temporal gait symmetry while utilizing all the provided biofeedback modes compared to no biofeedback ($P < 0.001$). Furthermore, two-way repeated measures ANOVA revealed statistically significant difference in symmetry ratio for main effect of vibration type (P-value = 0.016, partial eta squared = 0.585). The participants performed better with modes of biofeedback with varying vibration on-times. Furthermore, participants showed better performance when the biofeedback varied proportionally with the error. These findings suggest that biofeedback coding schemes may have a significant effect on the performance of gait training.**

***Index Terms*—Biofeedback, Gait Rehabilitation, Stroke, Symmetry, Vibrotactile.**

Manuscript received December 26, 2018; revised June 13, 2019; accepted June 18, 2019. Date of publication June 24, 2019; date of current version August 7, 2019. This research was supported in part by the National Research Foundation (NRF) of Korea under Grant 2019M3C1B8090798 and the Korea Evaluation Institute of Industrial Technology (KEIT) under Grant No. 20003822 and in part by Innovative Practical Use Project funded by the GIST in 2019. *(Corresponding authors: Min-Kyun Oh; Jungwon Yoon.)*

M. R. Afzal was with the School of Integrated Technology, Gwangju Institute of Science and Technology, Gwangju 61005, South Korea. He is now with the Department of Mechanical Engineering, KU Leuven, 3000 Leuven, Belgium (e-mail: raheel379@gmail.com).

H. Lee and J. Yoon are with the School of Integrated Technology, Gwangju Institute of Science and Technology, Gwangju 61005, South Korea (e-mail: eldoon@nate.com; jyoon@gist.ac.kr).

A. Eizad is with the School of Mechanical and Aerospace Engineering, Gyeongsang National University, Jinju 52828, South Korea (e-mail: amre.eizad@gmail.com).

C. H. Lee and M.-K. Oh are with the Department of Rehabilitation Medicine, Gyeongsang National University School of Medicine, Gyeongsang National University Hospital, Jinju 52727, South Korea (e-mail: ychkck1407@gmail.com; solioh21@hanmail.net).

Digital Object Identifier 10.1109/TNSRE.2019.2924682

I. INTRODUCTION

WORLDWIDE, stroke ranks as the leading cause of disability [1]. Lack of mobility due to post stroke gait abnormalities causes reduced quality of life and increased long-term care needs among stroke recovering individuals [2]. Thus, the primary goal of rehabilitation after stroke is to regain independent walking at home and in the community [3], [4]. Close supervision of a physical therapist is required to properly carry out the gait training procedures. Physical therapists address gait asymmetry to improve balance control and reduce gait inefficiencies. However, the patient to therapist ratio is becoming larger day by day, and creating a situation that demands the use of engineering technology to provide therapist like improvements in rehabilitation programs [5].

Gait rehabilitation is necessary to increase the functional walking ability of the patient to an acceptable level for performing normal tasks, thus reducing risk of subsequent health defects. A complexity in the achievement of this task is that the gait velocity and symmetry of individuals with mild to moderate stroke are affected by different physical impairments [7]. Therefore, due to the variations in the causes and symptoms of gait disorders, rehabilitative methods are often tailored specifically for the particular individual [8]. As task-specific and context-specific training are widely accepted principles for motor learning, the training scheme adopted for each individual should target their particular needs [37]. Adequate gait rehabilitation demands committed time with a therapist, expensive instrumentation and training devices, and the use of a gait laboratory. In order to reduce these complexities, technologies such as augmented feedback are being explored for use in rehabilitative procedures. Augmented feedback can play an important role when learning or improving a motor skill [9].

Vibrotactile systems have been used recently to provide feedback for postural sway reduction and gait rehabilitation applications. In community-dwelling elderly subjects, vibrotactile biofeedback enabled control of mediolateral sway during gait and reduction of fall risk [10]. Vibrotactile biofeedback reduced trunk sway during quiet standing and locomotor activities in individuals with vestibular loss [11], [12]. Balance training of individuals suffering from Parkinson's disease using a vibrotactile biofeedback system showed beneficial effects on trunk stability [13]. Likewise, use of vibrotactile biofeedback has been explored for gait training of individuals recovering from surgery to remediate knee osteoarthritis [14], athletes [15], transtibial amputees [16] and stroke survivors [17]. These studies demonstrate the effectiveness of vibrotactile biofeedback in providing instructional cueing. Hence, vibrotactors can provide effective cues with the least amount of interference to the subject's activities of daily life. The small size and low power consumption of these device elements makes them suitable for use in low-cost and easy-to-use portable biofeedback systems to provide overground gait training experience for gait rehabilitation after stroke. Vibrotactors have also garnered great interest due to the contemporary trend towards development of wearable systems. Recently, we developed a wearable system that provides vibrotactile cues for post-stroke gait training. Through pilot testing with that system, we identified the potential of vibrotactile biofeedback for gait symmetry training of hemiparetic stroke suffering individuals [18].

Different schemes of biofeedback delivery for postural control tasks have been studied in recent researches [19]–[21]. Dozza *et al.* found that, for postural control, sigmoid-coded audio biofeedback reduced sway acceleration more than a linear-coded audio biofeedback, whereas a linear-coded visual biofeedback reduced sway acceleration more than a sigmoid-coded visual biofeedback [19]. Kinnaird *et al.* found that in a standing postural control task performed by healthy older adults, performance improved with both attractive and repulsive cuing strategies. In each trial, the participants performed better when using repulsive cuing strategy; however, the rate of improvement over several trials was greater for attractive cuing [20]. Lee *et al.* found that, while performing dynamic exercises during an upright standing task, a continuous coding scheme produced significantly better performance than a binary scheme when elderly individuals with Parkinson's Disease and healthy elderly individuals performed dynamic weight-shifting balance exercises with assistive vibrotactile biofeedback [21]. Although, these studies provide a useful insight into biofeedback coding schemes for postural tasks, the application of such biofeedback coding schemes to gait symmetry training is not yet well explored. Hemiparesis results in reduced gait efficiency and reduced activity levels. This weakness or inability to properly use one side of the body causes gait asymmetry and increases risk of musculoskeletal injury to non-paretic limbs [30]. Therefore, in the presented research we have endeavored to explore the effect of different vibrotactile biofeedback coding schemes on the training of gait symmetry of individuals with hemiparetic stroke.

From our experimental study in [18], we acquired useful insight about the vibrotactile biofeedback system's performance and effects on gait symmetry. Based on that understanding, we modified the system hardware to improve its wearability and interface with PC [22]. In this paper, we have briefly described the design and components of our inexpensive wearable vibrotactile cue delivery system for gait training. The system description is followed by a description of the development of various biofeedback coding schemes. Afterwards, we have detailed the trials with individuals recovering from stroke through which we identified the efficacy of providing vibrotactile cues with different coding schemes for gait symmetry training.

II. MATERIALS AND METHODS

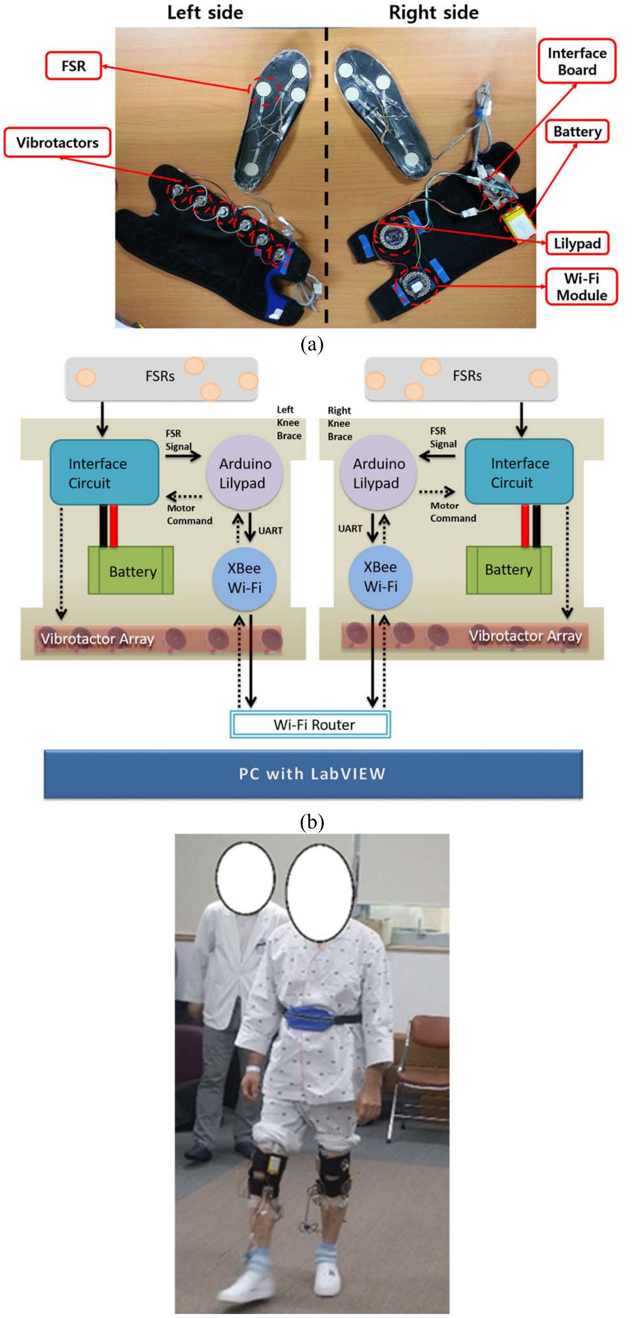
A. System

Individuals suffering from hemiparetic stroke commonly suffer from temporal asymmetry in gait cycle during ambulation [32]. To detect the temporal gait parameters such as stance-time and swing-time, contact based pressure-sensitive insoles can be utilized. Change in pressure applied to a point can be detected using Force Sensitive Resistor (FSR). Our system (Figure 1), can detect temporal gait symmetry to deliver various modes of vibrotactile biofeedback [23].

The insoles utilized in our system are designed with FSRs placed at Toe, Metatarsal 1, Metatarsal 5 and Heel. We have developed several insoles based on standard Korean shoe sizes, to accommodate different foot size of users. Each FSR (Tekscan, A401 [24]) can detect ground contact through adjustment of a threshold to distinguish between stance and swing phase of gait. Using this ground contact information important gait parameters like stance-time on each leg, swing-time on each leg and double-stance-time can be calculated. Manipulation of this time data is useful in determining gait abnormalities like temporal asymmetry. Each FSR is connected to an interface circuit with wires (one circuit for each leg); the circuit transforms the change in resistance to change in voltage. This change in voltage is thus used to determine the ground contact information. The interface circuit is connected to a Lilypad Arduino [25] (one Lilypad for each leg); the Lilypad continuously reads the analog voltage signals, converts them into 10bit digital values and sends this information to a Lilypad XBee module (one for each leg) via wired serial (UART) communication. XBee module uses a Wi-Fi transceiver [26] (one for each leg) to send this information to a PC running a custom-made program executing inside the LabVIEW (National Instruments) environment.

The FSR contact information received at the PC is run through the LabVIEW program to identify gait parameters and deliver vibrotactile biofeedback accordingly, at the same time providing monitoring and data storage functionality through the GUI. At each leg a 3.7V 2000mAh lithium polymer battery combined with a 3.3V voltage regulator provides the necessary power to run the portable system. The Lilypad Arduino, Wi-Fi XBee, Battery, Interface circuit and Vibrotactors are securely attached to soft knee wrap to make an easy to wear system.

At the PC the custom-made LabVIEW program receives the FSR ground contact data. This ground contact information is processed to obtain swing-time and stance-time of each leg of the subject. Furthermore, symmetry ratio (SR) is calculated for each gait cycle. SR is defined as the ratio between stance-time of healthy leg and stance-time of paretic leg [27]. Generally, the SR of a healthy individual will range between 0.95 and 1.05. Thus, an individual with stroke would exhibit a SR value of larger than 1.05 in case of an asymmetric gait pattern due to hemiparesis [28]. In this system, the vibrotactile biofeedback



(a)

(b)

(c)

Fig. 1. (a) The developed system hardware and its components (adapted from [23]). (b) Block diagram showing information flow. Each side has identical components and mirrored placements to allow its use with left and right hemiparetic stroke survivors (adapted from [22]). (c) Participants wore the two soft knee wraps upon which the whole system with its components is affixed and shoes containing the instrumented insoles.

signal is generated based on the error between a desired SR value and actual SR value.

The control signal for vibrotactile biofeedback is generated by the PC and is transmitted over Wi-Fi to be received by the Lilypad Arduino, which in turn switches the vibrotactor array (Precision Microdrives ™310-101 [29] on Lilypad-Vibe-Board, 6 for each leg) through a Darlington transistor. Operating at 3.3v, the amplitude of vibration is 1.5g at 100% intensity (with a test load of 100gram), and a rise time of 91ms. The vibrotactile feedback is delivered only at the paretic leg during the swing phase.

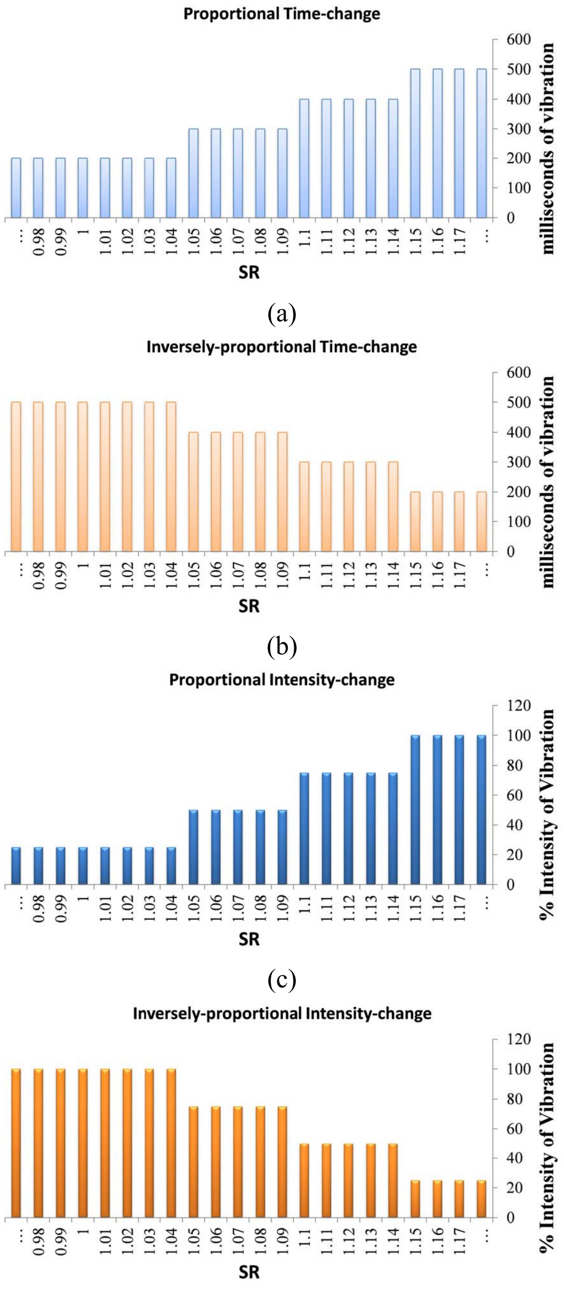
In order to devise the different feedback coding schemes, we looked at the different aspects of the processing of tactile stimuli. Human brain is prone to control motor tasks by processing and incorporating time-discrete somatosensory information in its internal models [33], [34]. So, we aim to identify the effects of a time variable feedback on gait symmetry training. Human motor performance and learning is affected by variable intensity biomechanical feedback [21], [35]. Thus, exploration of the effects of an intensity variable feedback on gait symmetry training is also warranted. Furthermore, we hypothesized that the user may like the increase in vibration as a sensory input signifying that they are doing the task correctly or they may consider it a nuisance that they would like to avoid by doing the task correctly. Therefore, the proportional and inversely-proportional relationship between the feedback and the error magnitude has also been considered. Hence, in this research we have employed four vibrotactile biofeedback modes (2 vibration types x 2 relation types) that are explained below.

1. Proportional Time-change (PT): vibration on-time is changed proportionally to error between desired and actual SR.
2. Inversely-proportional Time-change (IT): vibration on-time is changed inverse proportionally to error between desired and actual SR.
3. Proportional Intensity-change (PI): vibration intensity is changed proportionally to error between desired and actual SR.
4. Inversely-proportional Intensity-change (II): vibration intensity is changed inverse proportionally to error between desired and actual SR.

In all the modes, the change (on-time/intensity) is kept discrete to amplify the perception of tactile cues. In PT and IT modes, the vibration signal is provided at full intensity and only the on-time is changed between 500ms (maximum) and 200ms (minimum). In these modes, vibration starts with start of the swing phase of the paretic leg and ends at the end of the vibration on-time or the end of the swing phase, whichever comes first. In PI and II modes, the vibration signal remains on during the swing phase of the paretic leg and the intensity is changed between 100% (maximum) and 25% (minimum). The minimum intensity value was so selected to ensure that it was above the sensitivity threshold of the calf musculature [44] and that it was distinguishable from an unperceived feedback signal. Figure 2 further explains the modes of vibrotactile biofeedback used in this research, with an example of how an individual with stroke can make use of the biofeedback to achieve a target SR.

B. Participants

We conducted experimental trials involving eight individuals with stroke to test the system efficacy in providing vibrotactile cues for achieving a target SR. The purpose of doing this experiment was twofold; firstly, we wanted to quantify the effectiveness of each of the biofeedback modes in assisting the subjects to improve SR, and secondly, to ascertain the



(a)

(b)

(c)

(d)

Fig. 2. Modes of vibrotactile biofeedback ((a) PT, (b) IT, (c) PI and (d) II) depicted with assumed application for a subject with stroke whose objective is to reduce the asymmetric temporal gait pattern (target SR < 1.05). In all plots, horizontal axis represents the subject's symmetry ratio.

TABLE I

DEMOGRAPHIC DETAILS OF PARTICIPANTS

Demographic	Value
Participants	8
Age (year)	54.5 ± 6.1
Height (cm)	165.5 ± 8.7
Mass (kg)	67.6 ± 6.3
Days since onset	23.9 ± 9.3
Modified Barthel Index Score	75.0 ± 11.8
Mini-Mental State Examination Score	24.5 ± 4.1
Gender	Male=6, Female=2
Cause of Stroke	Infarction=5, Intracerebral Hemorrhage=3
Side of Hemiplegia	Right=1, Left=7

Values are mentioned as Mean ± SD or counts (as appropriate).

TABLE II

PROTOCOL OF EXPERIMENTAL TRIALS

Trial	Conditions
NB	Natural Walking with no feedback
PT	Walking with Proportional Time-change biofeedback for correction of asymmetry.
IT	Walking with Inversely-proportional Time-change biofeedback for correction of asymmetry.
PI	Walking with Proportional Intensity-change biofeedback for correction of asymmetry.
II	Walking with Inversely-proportional Intensity-change biofeedback for correction of asymmetry.

Two trials per condition were performed with a break of 2 min between iterations and 5 min between trials.

difference in achieved outcomes under various biofeedback modes. Participants of the study had suffered single onset of unilateral hemiparetic stroke, and were in the sub-acute phase of recovery. All participants were able to walk more than 10 m without assistance. Individuals who were unable to follow verbal requests, had foot drop condition, limitations in joint range of motion, pain in the lower extremities, unstable medical conditions, or other diagnosed neurologic or musculoskeletal diseases were excluded from the study. None of the subjects suffered from any peripheral neurological conditions. Details of the subjects who participated in this study are mentioned in Table 1.

C. Protocol

The trials commenced with introduction of the system and its use. Participants were given instructions on how to interpret the various biofeedback modes. The instrumented insoles were installed in their shoes and the wearable system that includes knee wraps was put on each of their legs and fastened using Velcro straps (see Figure 1). After this initial setup, each user walked on the ground to acquire threshold FSR values. These threshold values were subsequently used in the LabVIEW program to distinguish between stance and swing phase of each leg (see Figure 3.). The target SR was set at < 1.05 in order to compel the participants to correct their asymmetric gait. In order to get acquainted with the system, all subjects utilized the various modes of vibrotactile biofeedback prior to the recording of experimental trials.

Each subject underwent five trials of self-paced overground walking. All subjects underwent the trials in random order. The five trials were as follows: one trial of Normal Walk (NB), where the subject walked with normal gait without any feedback, and four trials with the biofeedback modes (PT, IT, PI, II) to achieve the target SR. Here, P stands for directly

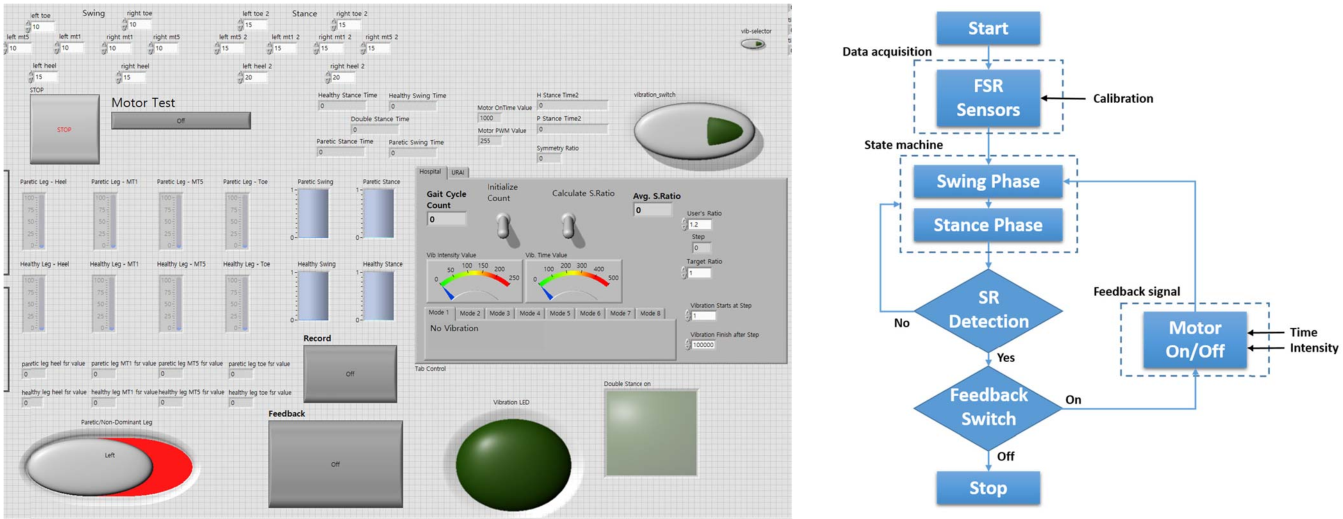


Fig. 3. (a) Graphical User Interface developed for the operation of the system in LabVIEW development environment. The operator has controls for calibrating the instrumented insoles, selecting the type of feedback and data recording. (b) Flowchart of the system control software. This diagram depicts a single iteration of the system workflow. In implementation, the modules work in parallel to ensure availability of data for each subsequent cycle.

TABLE III

THE MEAN OBSERVED SR DURING VARIOUS WALKING TRIALS

Subject	No Biofeedback (NB)	Proportional Time-change (PT)	Inv.-Proportional Time-change (IT)	Proportional Intensity-change (PI)	Inv.-Proportional Intensity-change (II)
1	1.1219	1.0464	1.0362	1.0987	1.0872
2	1.1472	1.0301	1.0323	1.0473	1.0689
3	1.1139	1.0605	1.0717	1.0594	1.0617
4	1.0978	1.0519	1.0795	1.0487	1.0918
5	1.1384	1.0553	1.0416	1.0771	1.0792
6	1.1436	1.0538	1.0609	1.0748	1.0897
7	1.1160	1.0506	1.0494	1.0577	1.0516
8	1.1377	1.0358	1.0469	1.0648	1.0674

SR is the ratio between stance-time of healthy leg and paretic leg. The custom-built LabVIEW program calculated the stance times of healthy and paretic leg. The values given here are mean SR of the participants during various modes of biofeedback.

proportional, I in the first letter space stands for inversely-proportional, T denotes variations in on-time of vibration and I in the second letter space denotes variations in intensity of the vibration. This study was conducted following the principles of the Declaration of Helsinki. All subjects gave written informed consent prior to data collection.

D. Data Analysis

The experimental trials were conducted with coordination of a PC operated by one of the researchers. The custom-built LabVIEW program handled two-way real-time data communication with the two Arduino Lilypads. The live-communication setup between LabVIEW and Arduinos was made possible with the use of XBEE Wi-Fi transceivers. The LabVIEW program monitored the signals from the FSRs and performed gait segmentation to determine the swing-time and stance-time at each leg (see Figure 4.). Subsequently, the ratio of stance-times was used to calculate SR. The LabVIEW program stored these SR values to determine the efficacy of vibrotactile biofeedback on gait symmetry of the participants. Gait speed during the various trials was measured using a handheld stopwatch by one of the researchers who observed the gait trials.

In post-experimental data analysis, these SR and gait speed values of the participants obtained during the five trials were evaluated. Using dependent t-test, we compared the SR and gait speed of trials done under different modes of feedback, here the statistical significance was defined at $P<0.05$. Furthermore, a two-way repeated measures ANOVA was used to identify the effects of vibration type (factor, 2 levels: time-change, intensity-change) and relation type (factor, 2 levels: proportional, inversely-proportional) on SR and gait speed while the subjects tried to achieve the target SR during various biofeedback modes. In addition, Mauchly's test of Sphericity was used to validate the repeated measures ANOVA. Post-hoc tests were conducted using the Bonferroni correction method. Partial eta squared was calculated as a measure of the effect size for two-way repeated measure ANOVA. Cohen's d was also calculated and used for pairwise comparisons. All statistical analysis was performed using SPSS V20.0 (IBM Corp., Armonk, NY, USA).

III. RESULTS

The mean observed SR and gait speed of each participant measured during the experiment are presented in Table 3 and 4, respectively.

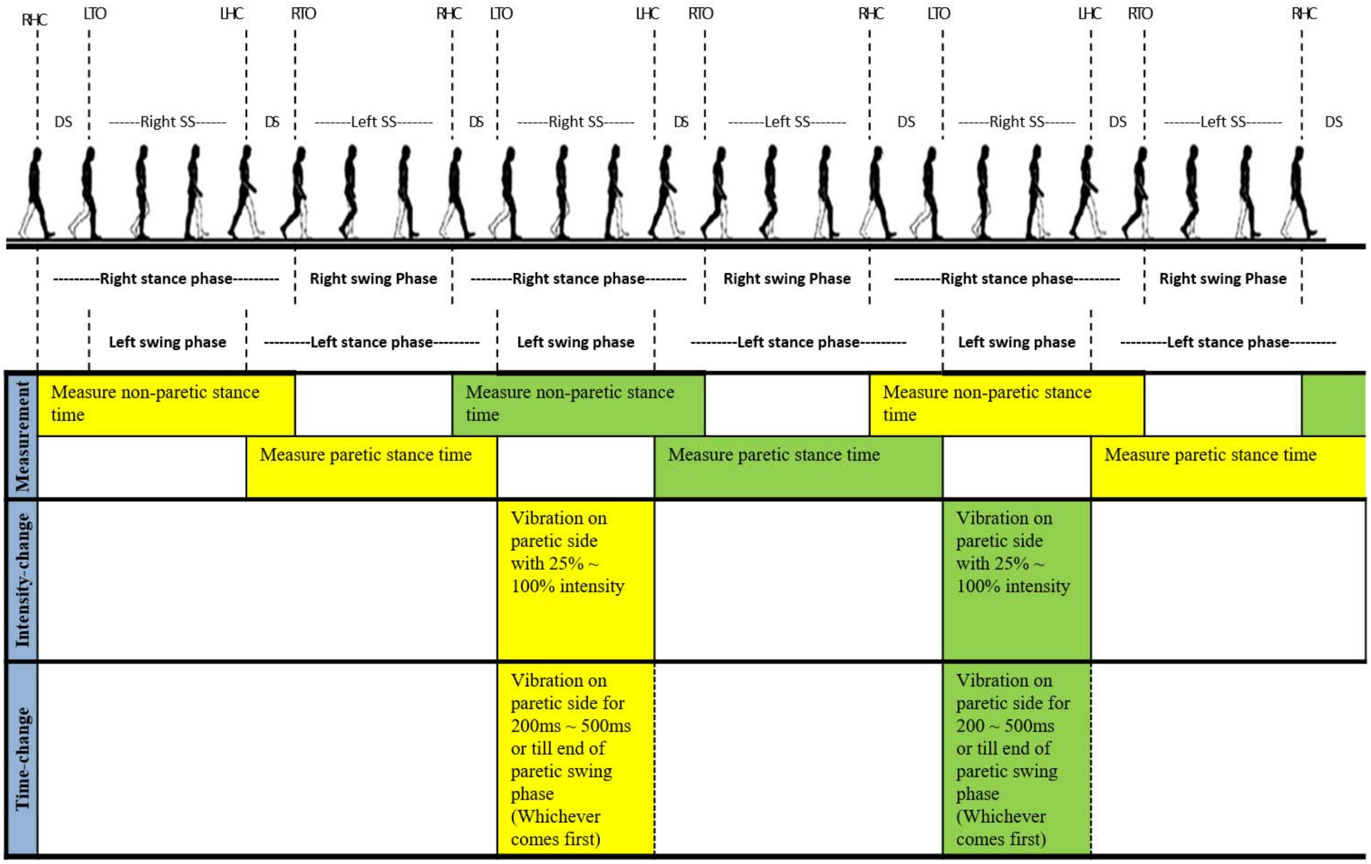


Fig. 4. Modes of vibrotactile biofeedback explained with subject's left-side assumed to be paretic (RHC: right heel contact, DS: double support, LTO: left toe off, Right SS: right single support, LHC: left heel contact, RTO: right toe off, and Left SS: left single support).

TABLE IV

THE MEAN OBSERVED GAIT SPEED DURING VARIOUS WALKING TRIALS

Subject	No Biofeedback (NB)	Proportional Time-change (PT)	Inv.-Proportional Time-change (IT)	Proportional Intensity-change (PI)	Inv.-Proportional Intensity-change (II)
1	0.49	0.63	0.57	0.70	0.68
2	0.77	0.75	0.86	0.83	0.89
3	0.76	0.69	0.71	0.68	0.65
4	0.77	0.63	0.66	0.69	0.61
5	0.59	0.54	0.68	0.45	0.60
6	0.68	0.55	0.61	0.62	0.61
7	0.81	0.76	0.72	0.80	0.78
8	0.52	0.60	0.57	0.65	0.63

Mean observed gait speeds (m/s) of each participant during various modes of biofeedback. Gait speed was measured using a handheld stopwatch by a researcher observing the trials.

The t-tests revealed statistically significant differences (P-value <0.001) between SR of NB trial and each of the biofeedback trials (PT, IT, PI and II) (see Figure 5). The Cohen's d values for pairwise comparisons of SR following the t-tests were as follows: NB vs PT = 5.55, NB vs. IT = 4.37, NB vs PI = 3.55, NB vs. II = 3.80. However, from t-tests, no statistically significant differences were observed between gait speeds of all trials (see Figure 6).

Results of two-way repeated measures ANOVA of SR and gait speed are presented in Table 5 and 6, respectively. There was a significant main effect of vibration type on the subjects' symmetry ratio (P-value = 0.016, partial eta squared = 0.585). No significant interaction between vibration type and relation type led to post-hoc analysis of relation types (proportional and inversely proportional) separately (see Table 5). Post-hoc analysis revealed significant difference between PT and PI (P-value = 0.025, Cohen's d = 1.28) and between IT and II (P-value = 0.017, Cohen's d = 1.42) (see Figure 7).

No statistically significant difference was found in gait speed between main effects of vibration type and relation type, and their interaction (see Table 6).

IV. DISCUSSION

In this paper, we presented the work we carried out using our wearable vibrotactile biofeedback system that is capable of providing various schemes of tactile cues for gait symmetry training. This system enables the user to experience over-ground walking while receiving vibrotactile biofeedback

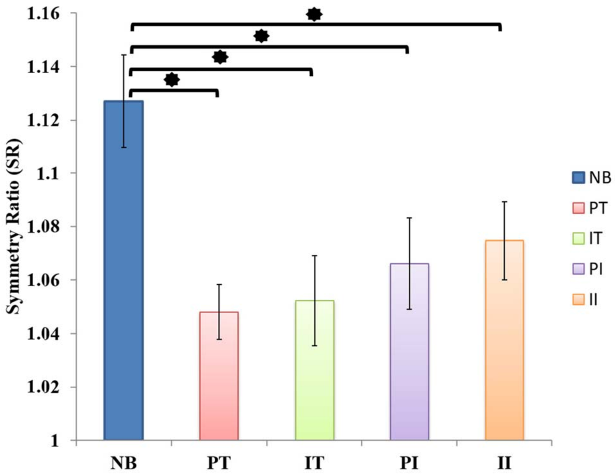


Fig. 5. Mean and SD of the SR of all participants during various walking trials, marked with statistically significant difference obtained from the t-tests. Here $*$ represents P-value < 0.001 and error bars represent the standard deviation.

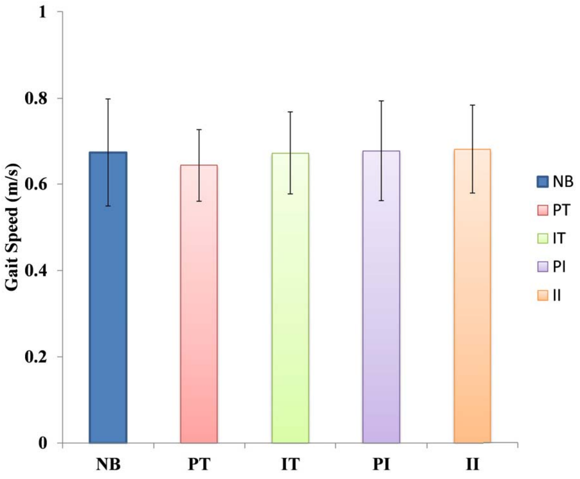


Fig. 6. Mean and SD of the gait speeds of all participants during various walking trials. T-tests revealed no statistically significant differences between them. Here, error bars represent the standard deviation.

TABLE V

STATISTICS OF THE TWO-WAY REPEATED MEASURES ANOVA FOR SR

	F	P-value	partial eta squared
Vibration Type (Time-Change vs. Intensity-Change)	**(1,7) = 9.857**	**0.016**	**0.585**
Relation Type (Prop. vs. Inv. - Prop.)	(1,7) = 1.618	0.244	0.188
Interaction	(1,7) = 1.119	0.325	0.138

Statistically significant p-value and subsequent effect size are indicated in bold.

TABLE VI

STATISTICS OF THE TWO-WAY REPEATED MEASURES ANOVA FOR GAIT SPEED

	F	P-value	partial eta squared
Vibration Type (Time-Change vs. Intensity Change)	(1,7) = 1.112	0.327	0.137
Relation Type (Prop. vs. Inv. Prop.)	(1,7) = 0.484	0.509	0.065
Interaction	(1,7) = 1.823	0.219	0.207

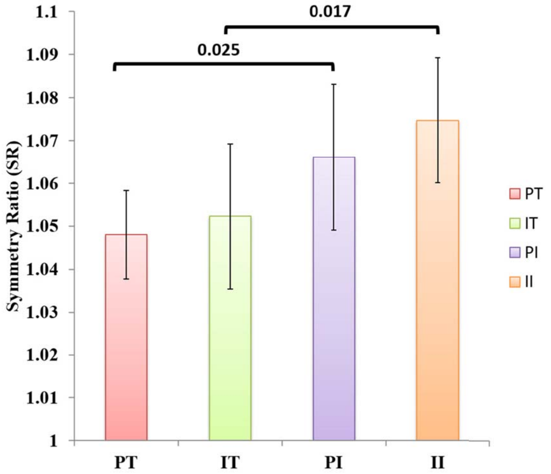


Fig. 7. Post-hoc analysis of SR under different biofeedback coding schemes. Analysis revealed significant difference between PT and PI (p = 0.025) and between IT and II (p = 0.017). Here, error bars represent the standard deviation.

for correction of temporal asymmetry of gait. The presented work focused on gauging the efficacy of various modes of vibrotactile feedback in ameliorating gait symmetry and on the comparison of the different modes in terms of the obtained outcomes. In this study, individuals with stroke underwent walking trials with the aim to achieve a target symmetry ratio while utilizing the provided modes of vibrotactile cues. These modes were defined by changes in vibration on-time or intensity, and their relation to error between desired and actual SR, i.e. the changes were either proportionally or inverse proportionally related to the magnitude of the error. All subjects were able to utilize the various coding schemes of vibrotactile biofeedback. The walking trials conducted under these different biofeedback conditions allowed us to observe the differences in performance of the subjects in achieving a target symmetry ratio. The results show that coding schemes can potentially have a significant effect on the training of gait symmetry.

The participants of this study were individuals recovering from hemiparetic stroke and were in the gait rehabilitation phase of recovery. Hemiparesis, or one-sided weakness, is the most common movement impairment among stroke survivors. The most efficient gait pattern is a symmetrical one, as gait asymmetry and energy expenditure are correlated [31]. In post-stroke chronic stage, 55.5% of the individuals exhibit temporal asymmetry [32]. Post-stroke stance-time symmetry ratio is correlated with swing-time and step-length symmetry ratio, thus it can indicate improvements in gait symmetry [27], [28]. There exists a possibility that training using the split-belt treadmill may remediate post-stroke asymmetric walking patterns [6]. However, it is not easy for the individuals requiring rehabilitation to use the split-belt treadmill rehabilitation systems by themselves on a daily basis. This difficulty is caused in part by the large size and complex working of the equipment involved. Another cause of difficulty is the high cost of such equipment that hinders its wide availability. Hence, there is an evident need for a low-cost in-home usable system for gait rehabilitation using biofeedback.

The participants of the study were able to achieve the target SR with all the biofeedback modes. As compared to the no biofeedback walking trial, all biofeedback walking trials exhibited statistically significant improvements in SR of the participants. However, the participants performed better with the biofeedback modes where the vibration on-time varied with the error in SR. The outcome of two-way repeated measures ANOVA revealed statistically significant difference (P-value = 0.016, partial eta squared = 0.585) between time encoded and intensity encoded vibrotactile biofeedback. Post-hoc analysis revealed significant difference between PT and PI (P-value = 0.025, Cohen's d = 1.28) and between IT and II (P-value = 0.017, Cohen's d = 1.42). Time-discrete somatosensory cues lead to the perception of a rhythm that the subjects can incorporate in their body control scheme and learn to associate with a physiological gait pattern, without having to pay continuous attention to the modulation of a time-continuous stimulation [36]. The current study shows that it is plausible that such time-discrete stimuli provided during cyclical tasks may provide a rhythm that can be integrated in the body control scheme and hence improve locomotion [16].

Gait speed of the participants during various modes of biofeedback was also observed (see Table 4). Gait speed is considered to be an important marker of deficit severity and functional ability after stroke [42]. Gait speed is adversely affected by increase in cognitive load [43]. We observed no statistically significant difference in participant gait speeds during the various biofeedback modes. This maintenance of gait speed was observed among all participants, showing that the various modes of biofeedback may not have imposed a high cognitive load.

Differences between how the central nervous system naturally processes tactile input and the afferent information of on-time or intensity may explain why the different modes of biofeedback generated different outcomes. Perceptual thresholds for haptic afferents depend on type of skin, stimulus type, and timing of the stimulus [38]. Mechanoreceptors are responsible for the perception of stimulus intensity [40], whereas different parts of the human brain, including cerebellum, cerebral cortices and basal ganglia, cater to the temporal perception of short-term events [39]. As somatosensory sensitivity in subjects with stroke is compromised, their perception of afferent intensity variations may be reduced [41]. While subjects can use these different modes of biofeedback to achieve the target, the outcomes may also differ due to user preference and/or ease-of-understanding of certain schemes. We initially hypothesized that the subjects would prefer vibrotactile biofeedback with either proportional or inversely-proportional relationship with error magnitude. The results show no statistically significant difference in SR between the two relationships. However, the subjects did perform better with the proportional relationship (increase in error causes increase in feedback magnitude) over the inversely-proportional one (increase in error causes decrease in feedback magnitude). The difference between observed values of SR during PT (1.0481 ± 0.0103) and IT (1.0523 ± 0.0169) trials, and that between PI (1.0661 ± 0.0170) and II (1.0747 ± 0.0146) trials highlights this phenomenon. This may be attributed to the general preference that users have for intuitive feedback schemes [38].

Limitations of the present study are the relatively small sample size, and an imbalance in gender distribution and side of hemiplegia. Despite the limitations, this study suggests that biofeedback schemes may have a significant effect on the performance of gait training. Therefore, careful selection of the biofeedback scheme is necessary.

The eventual goal of this research is to devise a system that can be used in non-laboratory settings for gait rehabilitation of individuals with sub-acute stroke. The modifications necessary to achieve that include the use of a smartphone based interface to replace the PC used in the current system. Future work includes application of this system with its various vibrotactile biofeedback coding schemes for gait symmetry training and identification of the mode preferred by individuals suffering from gait asymmetry due to ailments other than stroke. In addition, the changes in lower-limb kinematics corresponding to the use of various biofeedback schemes warrant further investigation.

V. CONCLUSION

In the current research, we analyzed the efficacy of various biofeedback modes provided to individuals recovering from stroke tasked with improving their temporal gait symmetry. The different vibrotactile biofeedback modes used included variations of on-time or intensity with varying relationships with error magnitude. We carried out walking trials of individuals with stroke to determine the effectiveness of the different biofeedback modes in assisting the subjects to improve SR, and to ascertain the difference among outcomes achieved with them. Results show that while using all the biofeedback modes, subjects were able to significantly improve their SR. Upon comparison of outcomes obtained with the different modes, we found that the subjects preferred feedback provided in terms of vibration on-time variation as compared to that with intensity variation. We also found that the relationship between feedback and error magnitude influences performance of the subjects. The results of the current study suggest that the particular biofeedback scheme chosen may have a significant effect on the performance of gait training. This may be due to neurophysiological factors and/or a preference that the subject may have. To enable the inclusion of vibrotactile coding schemes in post-stroke gait symmetry training it is necessary to identify the underlying mechanism that causes varied performance among subjects. Future researches should also investigate the extent of differences in performance among a more diverse (severity, type) group of individuals with stroke.

ACKNOWLEDGEMENTS

M.-K. Oh managed the patient trials and contributed equally to this work as a corresponding author.

REFERENCES

[1] J. Adamson, A. Beswick, and S. Ebrahim, "Is stroke the most common cause of disability?" *J. Stroke Cerebrovascular Diseases*, vol. 13, no. 4, pp. 171–177, Jul. 2004.

[2] N. E. Mayo, S. Wood-Dauphinee, R. Côté, L. Durcan, and J. Carlton, "Activity, participation, and quality of life 6 months post-stroke," *Arch. Phys. Med. Rehabil.*, vol. 83, no. 8, pp. 1035–1042, Aug. 2002.

[3] S. E. Lord, K. McPherson, H. K. McNaughton, L. Rochester, and M. Weatherall, "Community ambulation after stroke: How important and obtainable is it and what measures appear predictive?" *Arch. Phys. Med. Rehabil.*, vol. 85, no. 2, pp. 234–239, Feb. 2004.

[4] S. E. Lord and L. Rochester, "Measurement of community ambulation after stroke," *Stroke*, vol. 36, no. 7, pp. 1457–1461, Jul. 2005.

[5] G. McHugh and I. D. Swain, "A comparison between reported and ideal patient-to-therapist ratios for stroke rehabilitation," *Health*, vol. 05, no. 6, pp. 105–112, Jun. 2013.

[6] D. S. Reisman, R. Wityk, K. Silver, and A. J. Bastian, "Locomotor adaptation on a split-belt treadmill can improve walking symmetry post-stroke," *Brain*, vol. 130, no. 7, pp. 1861–1872, Jul. 2007.

[7] A.-L. Hsu, P.-F. Tang, and M.-H. Jan, "Analysis of impairments influencing gait velocity and asymmetry of hemiplegic patients after mild to moderate stroke," *Arch. Phys. Med. Rehabil.*, vol. 84, no. 8, pp. 1185–1193, Aug. 2003.

[8] F. A. Rubino, "Gait disorders," *Neurologist*, vol. 8, no. 4, pp. 254–262, Jul. 2002.

[9] B. Lauber and M. Keller, "Improving motor performance: Selected aspects of augmented feedback in exercise and health," *Eur. J. Sport Sci.*, vol. 14, no. 1, pp. 36–43, 2012.

[10] C. Wall, D. M. Wrisley, and K. D. Statler, "Vibrotactile tilt feedback improves dynamic gait index: A fall risk indicator in older adults," *Gait Posture*, vol. 30, no. 1, pp. 16–21, Jul. 2009.

[11] K. H. Sienko, M. Balkwill, L. I. E. Oddsson, and C. Wall, "The effect of vibrotactile feedback on postural sway during locomotor activities," *J. NeuroEng. Rehabil.*, vol. 10, no. 1, p. 93, Jul. 2013.

[12] B.-C. Lee, J. Kim, S. Chen, and K. H. Sienko, "Cell phone based balance trainer," *J. Neuroeng. Rehabil.*, vol. 9, no. 1, p. 10, 2012.

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

[14] P. Shull, K. Lurie, M. Shin, T. Besier, and M. Cutkosky, "Haptic gait retraining for knee osteoarthritis treatment," in *Proc. IEEE Haptics Symp.*, Mar. 2010, pp. 409–416.

[15] M. Fiorentino, A. E. Uva, and M. M. Foglia, "Self calibrating wearable active running asymmetry measurement and correction," *J. Control Eng. Appl. Inform.*, vol. 13, no. 2, pp. 3–8, Jun. 2011.

[16] S. Crea, B. B. Edin, K. Knaepen, R. Meeusen, and N. Vitiello, "Time-discrete vibrotactile feedback contributes to improved gait symmetry in patients with lower limb amputations: Case series," *Phys. Therapy*, vol. 97, no. 2, pp. 198–207, Feb. 2016.

[17] C. Z. Ma, Y.-P. Zheng, and W. C. Lee, "Changes in gait and plantar foot loading upon using vibrotactile wearable biofeedback system in patients with stroke," *Topics Stroke Rehabil.*, vol. 25, no. 1, pp. 20–27, Jan. 2018.

[18] M. R. Afzal, M.-K. Oh, C.-H. Lee, Y. S. Park, and J. Yoon, "A portable gait asymmetry rehabilitation system for individuals with stroke using a vibrotactile feedback," *BioMed. Res. Int.*, vol. 2015, Apr. 2015, Art. no. 375638.

[19] M. Dozza, L. Chiari, F. Hlavacka, A. Cappello, and F. B. Horak, "Effects of linear versus sigmoid coding of visual or audio biofeedback for the control of upright stance," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 14, no. 4, pp. 505–512, Dec. 2006.

[20] C. Kinnaird, J. Lee, W. J. Carender, M. Kabeto, B. Martin, and K. H. Sienko, "The effects of attractive vs. repulsive instructional cuing on balance performance," *J. NeuroEng. Rehabil.*, vol. 13, no. 1, p. 29, Mar. 2016.

[21] B.-C. Lee, A. Fung, and T. Thrasher, "The effects of coding schemes on vibrotactile biofeedback for dynamic balance training in Parkinson's Disease and healthy elderly individuals," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 26, no. 1, pp. 153–160, Jan. 2018.

[22] M. R. Afzal, H. Lee, J. Yoon, M.-K. Oh, and C.-H. Lee, "Development of an augmented feedback system for training of gait improvement using vibrotactile cues," in *Proc. 14th Int. Conf. Ubiquitous Robots Ambient Intell. (URAI)*, Jun./Jul. 2017, pp. 818–823.

[23] M. R. Afzal, H. Lee, A. Eizad, C. H. Lee, M.-K. Oh, and J. Yoon, "Evaluation of novel vibrotactile biofeedback coding schemes for gait symmetry training," in *Proc. 2nd IEEE Int. Conf. Soft Robot. (RoboSoft)*, Apr. 2019, pp. 540–545.

[24] Tekscan. *FSR A401*. Accessed: Nov. 5, 2018. [Online]. Available: https://www.tekscan.com/products-solutions/force-sensors/a401

[25] Arduino. *Lilypad Arduino Main Board*. Accessed: Nov. 5, 2018. [Online]. Available: https://store.arduino.cc/usa/lilypad-arduino-main-board

[26] Digi. *XBee Wi-Fi*. Accessed: Nov. 5, 2018. [Online]. Available: https://www.digi.com/products/xbee-rf-solutions/2-4-ghz-modules/xbee-wi-fi

[27] K. K. Patterson, W. H. Gage, D. Brooks, S. E. Black, and W. E. McIlroy, "Evaluation of gait symmetry after stroke: A comparison of current methods and recommendations for standardization," *Gait Posture*, vol. 31, no. 2, pp. 241–246, 2010.

[28] K. K. Patterson, A. Mansfield, L. Biasin, K. Brunton, E. L. Inness, and W. E. McIlroy, "Longitudinal changes in poststroke spatiotemporal gait asymmetry over inpatient rehabilitation," *Neurorehabil. Neural Repair*, vol. 29, no. 2, pp. 153–162, Feb. 2015.

[29] *Precision Microdrives, Model # 310-101*. Accessed: Nov. 5, 2018. [Online]. Available: https://www.precisionmicrodrives.com/wp-content/uploads/2016/04/310-101-datasheet.pdf

[30] I. G. van de Port, G. Kwakkel, V. P. Schepers, and E. Lindeman, "Predicting mobility outcome one year after stroke: A prospective cohort study," *J. Rehabil. Med.*, vol. 38, no. 4, pp. 218–223, Jul. 2006.

[31] M. M. Platts, D. Rafferty, and L. Paul, "Metabolic cost of over ground gait in younger stroke patients and healthy controls," *Med. Sci. Sports Exerc.*, vol. 38, no. 6, pp. 1041–1046, Jun. 2006.

[32] K. K. Patterson *et al.*, "Gait asymmetry in community-ambulating stroke survivors," *Arch. Phys. Med. Rehabil.*, vol. 89, no. 2, pp. 304–310, Feb. 2008.

[33] C. Cipriani, J. L. Segil, F. Clemente, R. F. ff. Weir, and B. Edin, "Humans can integrate feedback of discrete events in their sensorimotor control of a robotic hand," *Exp. Brain Res.*, vol. 232, no. 11, pp. 3421–3429, Nov. 2014.

[34] S. Crea, C. Cipriani, M. Donati, M. C. Carrozza, and N. Vitiello, "Providing time-discrete gait information by wearable feedback apparatus for lower-limb amputees: Usability and functional validation," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 23, no. 2, pp. 250–257, Mar. 2015.

[35] J. Lieberman and C. Breazeal, "TIKL: Development of a wearable vibrotactile feedback suit for improved human motor learning," *IEEE Trans. Robot.*, vol. 23, no. 5, pp. 919–926, Oct. 2007.

[36] M. Roerdink, C. J. Lamoth, G. Kwakkel, P. C. van Wieringen, and P. J. Beek, "Gait coordination after stroke: Benefits of acoustically paced treadmill walking," *Phys. Therapy*, vol. 87, no. 8, pp. 1009–1022, Aug. 2007.

[37] P. Langhorne, J. Bernhardt, and G. Kwakkel, "Stroke rehabilitation," *Lancet*, vol. 377, no. 9778, pp. 1693–1702, 2011.

[38] K. S. Hale and K. M. Stanney, "Deriving haptic design guidelines from human physiological, psychophysical, and neurological foundations," *IEEE Comput. Graph. Appl.*, vol. 24, no. 2, pp. 33–39, Mar./Apr. 2004.

[39] S. Grondin, "Timing and time perception: A review of recent behavioral and neuroscience findings and theoretical directions," *Attention, Perception, Psychophys.*, vol. 72, no. 3, pp. 561–582, Apr. 2010.

[40] A. Chatterjee, V. Aggarwal, A. Ramos, S. Acharya, and N. V. Thakor, "A brain-computer interface with vibrotactile biofeedback for haptic information," *J. NeuroEng. Rehabil.*, vol. 4, no. 1, p. 40, Oct. 2007.

[41] L. M. Carey, T. A. Matyas, and L. E. Oke, "Sensory loss in stroke patients: Effective training of tactile and proprioceptive discrimination," *Arch. Phys. Med. Rehabil.*, vol. 74, no. 6, pp. 602–611, Jun. 1993.

[42] S. L. Patterson *et al.*, "Determinants of walking function after stroke: Differences by deficit severity," *Arch. Phys. Med. Rehabil.*, vol. 88, no. 1, pp. 115–119, Jan. 2007.

[43] Y.-R. Yang, Y.-C. Chen, C.-S. Lee, S.-J. Cheng, and R.-Y. Wang, "Dual-task-related gait changes in individuals with stroke," *Gait Posture*, vol. 25, no. 2, pp. 185–190, Feb. 2007.

[44] A. Wilska, "On the vibrational sensitivity in different regions of the body surface," *Acta Physiologica Scandinavica*, vol. 31, nos. 2–3, pp. 285–289, Jul. 1954.