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Use of Vibrotactile Bracelets to Study Effects of Arm Swing Variation on Overground Gait

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This work involved human subjects in its research. Approval of all ethical and experimental procedures and protocols was granted by the Bioethics Review Committee of Gwangju Institute of Science and Technology, South Korea under Application No. 20201008-HR-56-03-04, and performed in line with the Declaration of Helsinki.

ABSTRACT The high portability of vibrotactile feedback systems makes them suited to wearable applications, which improves their usability for rehabilitation applications encompassing a variety of environments and scenarios. A number of works have explored the relationship between arm movement and gait parameters such as gait variations and age on the arm swing. However, the inter-limb coupling scheme, i.e. the effects of the specific side (left or right) and direction (forward or backward) of arm swing variation on gait and balance parameters have not yet been evaluated. The study of these effects can enable us to devise arm movement based gait training protocols that may be beneficial for stroke survivors. We have developed a vibrotactile biofeedback system worn on the upper limb for post-stroke gait rehabilitation training. Using this system, we have carried out a study with ten healthy subjects and one stroke survivor to determine the effects of arm swing variation on gait and balance parameters. The healthy subject experiments revealed that increase in arm swing significantly increases the stride length while bringing about a statistically non-significant increase in the gait velocity. The study also revealed that the protocols involving variation of forward arm swing appear to have greater efficacy in modifying the gait symmetry ratio. Furthermore, the variations in arm swing and the resulting gait modifications do not produce any significant difference in the balance parameters. The results from the pilot test with one stroke survivor also show that increasing the arm swing increases the stride length and velocity. These findings suggest that arm swing variation using vibrotactile bracelets has effects on gait parameters that may be utilized for gait training of stroke survivors.

INDEX TERMS Gait rehabilitation, motion measurement, rehabilitation robotics, vibration feedback.

I. INTRODUCTION

Worldwide, stroke is considered a major cause of long-term disability [1]. The after-effects of stroke usually decrease the mobility of survivors, which makes their community life difficult and reduces their quality of life. Therefore, the ultimate goal of gait rehabilitation for stroke survivors is to enable them to return to a better quality of community life [2], [3]. A common chronic effect of stroke is partial or one-sided paralysis that may cause slow and asymmetric gait with decreased dynamic stability [4]–[6]. In a clinical setting, therapists assist the patients in restoring their gait function. However, once a patient is discharged from the hospital, maintaining or further improving their gait ability becomes a challenge. Thus, making it difficult for stroke survivors to adapt to community life after leaving the hospital.

Majority of the commercialized robotic devices for gait rehabilitation suffer from limited use due to their high cost, dedicated space requirement and requirement of specialized training for the operators [7]. Furthermore, these devices show better results than conventional therapy with people who are in the acute and subacute stages of recovery from

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stroke, while the effects on chronic subjects do not differ from conventional therapy [8]. On the other hand, biofeedback based wearable systems can be made to have high flexibility for use by different types of users under different gait scenarios [6].

Biofeedback based systems assist the users by guiding their movements in order to achieve the training goals. Biofeedback has been reported to provide good motor function recovery when used for task-oriented training [9]. Biofeedback can be delivered through a number of different modalities such as auditory, visual, tactile, etc. Vibrotactile feedback generates only minimal interference to the user's activities of daily life [10]. Moreover, vibrotactile feedback systems can be manufactured at a low cost and can be made highly portable so that they can be used in the overground walking scenario [10]. Such systems, designed to deliver vibrotactile biofeedback to the user's calves, have been shown to provide promising results in terms of improving gait symmetry of stroke survivors [10], [11]. Furthermore, vibrotactile biofeedback was confirmed to improve the dynamic gait index and the control of mediolateral sway during gait [12]. The demonstrated efficacy and the possibility of implementation as a wearable package improves the usability of vibrotactile feedback systems for rehabilitation applications encompassing a variety of environments and scenarios.

Walking is a rhythmic movement that involves the coupling of upper and lower limbs [13]. The swing of the arm causes the generation of a moment about the vertical axis that serves to reduce the reaction moment generated at the foot in the stance phase for locomotion of the body [14]. Thus, arm swing serves to minimize the energy expenditure and to optimize the stability [15]. The range of arm swing has been shown to increase with the increase in walking speed [16], [17]. It has also been shown that the amplitude of arm swing increases with the increase in stride length and vice versa [18]. Due to this close relationship between arm swing and gait parameters, the modification of arm swing as a means of achieving gait rehabilitation may hold potential [15]. Some interesting researches related to this topic have reported that after instruction, stroke survivors could increase their arm swing on both sides during both in-phase (same) and out-of-phase (reverse) synchronization between the upper and lower limbs [19], and that under the effects of auditory rhythms, they could increase both their arm swing and stride length [20]. Note that both these researches were carried out under controlled speed conditions. It has also been reported that arm cycling of stroke patients sitting in a custom adapted wheel chair reduced the hyperactive soleus H-reflexes [21]. Furthermore, Stephenson *et al.* showed that stroke patients' arm movement to support their weight using horizontal freely moving parallel bars caused a decrease in the asymmetric activation of their lower limb muscles [22].

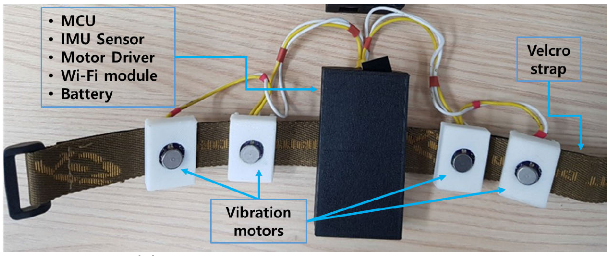
In order to explore the potential benefits of using arm swing modification as a means of gait rehabilitation, we first need an evaluation of the effects of arm swing modifications on the gait parameters. However, until now, most of the studies have focused on the evaluation of the effects of factors such as gait variations and age on arm swing [16], [[17], [23], [24]. In rehabilitation related works, a recent study with Parkinson's Disease patients has reported that vibration feedback aimed at increasing the swing amplitude of both arms at comfortable and fast paced gaits has shown promising results in terms of trunk sway reduction [25]. Another study investigated that addition of light arm weights (0.45 kg) affects the gait parameters (gait speed, cadence or stride length etc.) of Parkinsonian and healthy subjects [26]. It should be noted that the aforementioned studies on the role of upper extremities in walking of stroke patients have been conducted with upper and lower extremity synchronization either through instruction or through the use of an arm cycling device [19]–[22]. To the best of our knowledge, studies evaluating the effects of side specific (left or right) and direction specific (forward or backward) arm swing variations on gait parameters such as stride length and symmetry have not yet been reported. Increasing gait speed and symmetry is important for stroke patients as their increase is associated with an increase in gait stability and the ability of the stroke survivor to reintegrate in to community living [11]. Thus, our first hypothesis is that when a human increases their arm swing to more than that during normal walking, the stride length and gait speed also will be increased. Furthermore, [13] reported that correlations between diagonal upper/lower limb trajectories increased as the right-side belt speed of a split belt treadmill was increased, suggesting an increase in cross-body matching regardless of side. This leads to our second hypothesis that increase in the arm swing amplitude of one side will affect the step length of the related lower limb, which may be utilized to modify gait symmetry.

Building upon the abovementioned knowledge, we have developed an upper limb wearable vibrotactile biofeedback system for post-stroke gait rehabilitation training [27]. However, in order to develop an effective strategy for stroke rehabilitation, we first need to verify our hypotheses and identify the biofeedback strategies that are effective with the healthy subjects. Therefore, in this work we have utilized our developed system [27] to implement multiple biofeedback strategies to help the healthy subjects achieve different prescribed arm swing modifications. The effects of these modifications on the healthy subjects' gait have been analyzed to reveal the best performing biofeedback strategies. In addition, to check whether the protocol disturbs the subjects' gait balance, their body tilts during the tests are recorded for post-experimental analysis. Furthermore, the strategies thus selected are applied to a stroke survivor in a pilot trial to determine whether or not they can respond to the protocol i.e. they can modify their arm swing and that the modification brings a change in gait parameters.

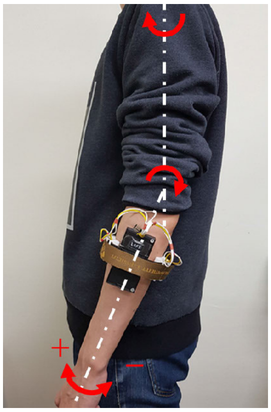
II. MATERIALS AND METHODS

A. MOVEMENT TRACKING AND BIOFEEDBACK SYSTEM

The developed haptic bracelet is shown in Fig. 1. It consists of four vibration motors (310-101 10mm vibration motor,



(a) The developed vibrotactile bracelet.



(b) The arm swing configuration while wearing the haptic bracelet

FIGURE 1. The movement tracking and vibrotactile biofeedback system.

Precision Micro-drives, UK, installed on the Lilypad Vibe board, Sparkfun, USA), an inertial measurement unit (IMU), a microcontroller (MCU), a motor driver and a Wi-Fi module. All the bracelet hardware including an 850mAh lithium-ion battery are enclosed in 3D printed plastic casings that are threaded onto a Velcro® covered strap that can be wound around the user's forearm as shown Fig 1 (b). The details of the bracelet's constituent parts are given in Table 1. The overall system is designed to operate with two bracelets, worn one on each arm. The combined weight of the two bracelets is 210g. The maximum vibration intensity of the vibration motors is 0.8G [28], and with all the motors are running continuously, the system has a calculated endurance of over two hours on one full charge of the onboard batteries. As shown in fig. 1 (a), the bracelet components are housed in separate casings (called blocks). The four small blocks house one vibration motor each and the rest of the system hardware is housed in the large block (43 × 82 × 25 mm) at the center of the bracelet.

The perception of vibration deteriorates due to movements [29]. Therefore, in this research, three strategies have been employed to ensure high level of vibration perception. First, the vibrotactile blocks are designed to have the vibration motors protrude about 2mm from the block surface. This has been done to increase the vibration perception by making the vibration motors to come in direct contact with the user's skin. Second, according to Oakley et al, it is possible to increase the intensity of vibration by using several vibration motors [30]. Therefore, four small and flat coin-type vibration motors are used in the presented wearable system. The motors are individually controllable to allow use of the device

TABLE 1. Details of system components.

Name	Details
IMU sensor	MPU9250, Invensense, USA
Microcontroller	SAMD21, Atmel, USA
Mainboard	9DoF Razor IMU M0, Sparkfun, USA
Vibration motor	310-101, Precision Microdrives, UK
Motor Driver	ULN2803A, Toshiba, Japan
Wi-Fi module	ESP8266, Espressif Systems, China
Battery	3.7V 850mAh Lithium ion battery

for future research applications such as haptic rendering and directional cuing. Third, we had the subjects wear the system on the upper forearm as it has more muscle and larger surface area than the wrist, which provides greater contact area between the arm and the bracelet, thus allowing us to use all the four vibration motors to achieve higher vibration recognition.

As shown in the block diagram (Fig. 2), the MCU and the IMU are both embedded on one mainboard. Sensor management, communication with the laptop and generation of control signals for the vibration motors, are all handled by the MCU. The MCU reads the sensor data and communicates with the laptop via the Wi-Fi module at a rate of 50 Hz. At the laptop, a software running in the LabVIEW® environment uses the sensor data to determine the magnitude of arm swing. This software also generates vibration commands based on the arm swing value and feedback strategy, and communicates them to the MCU via the Wi-Fi link. The MCU then generates the drive signals for the motor driver according to the received vibration commands. The laptop based software also stores the arm swing data for further analysis.

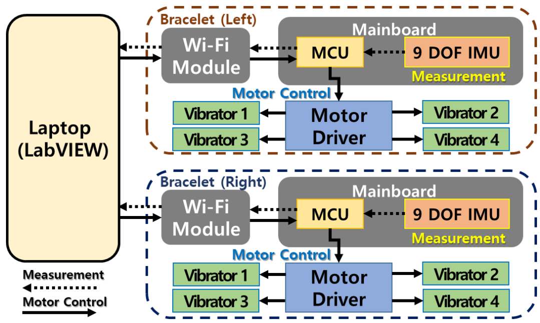


FIGURE 2. Operational block diagram of the entire system.

The developed vibrotactile bracelets determine the arm swing angle by calculating the overall arm movement in the sagittal plane irrespective of the shoulder flexion, rotation and abduction that occur during walking. The detailed method of this calculation is given in [27]. In addition, in the previous study [27], we verified the usefulness of the real-time arm angle measurement capability of the vibrotactile bracelets through comparison with an optical tracking system.

B. PARTICIPANTS

The study presented here was carried out in two parts. In the first part, trials with 10 healthy young subjects were conducted to verify our hypotheses, and to determine the suitability of the experimental protocol for stroke survivors by checking if the modification of arm swing has a significant effect on the subjects' balance (mediolateral, anteroposterior tilt of pelvis) during walking. In the second part, the best performing strategies from the first part were pilot tested with one stroke survivor and their effects were analyzed. The testing protocols used for the healthy subject and stroke subject studies are detailed in the A subsections of sections III and IV, respectively.

1) HEALTHY SUBJECTS

The ten young healthy subjects (Gender: 8 Males and 2 Females, Age: 28.9 ± 5.7 years, Height: 171.1 ± 6.2 cm, and Weight: 73.9 ± 12.3 kg, Dominant arm: 10 Right) who took part in this study did not suffer from any physiological disorders that might affect their gait or their ability to perceive and process vibrotactile cues.

2) STROKE PATIENT

One subject with stroke, who was capable of independent gait took part in this pilot study. Fig. 3 shows the test environment for this study and the demographic details of the subject are given in Table 2.

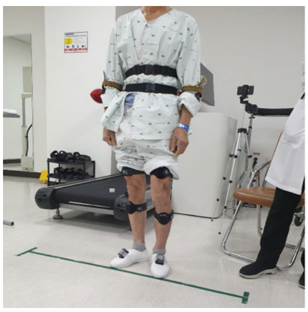


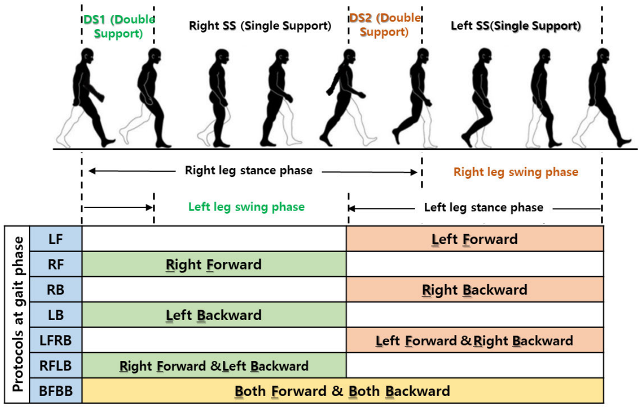
FIGURE 3. Experiment environment for the stroke subject test.

TABLE 2. Demographic details of the stroke subject.

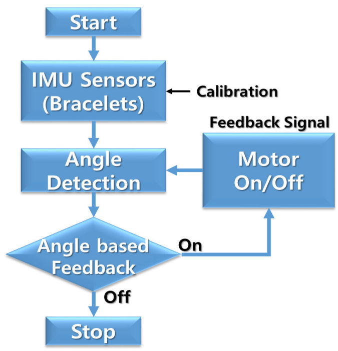
Gender	Age (year)	Height (cm)	Mass (kg)	Days since onset	Side of Hemi plegia	Cause of Stroke	Modified Barthel Index Score	Mini-Mental State Exam. Score
Male	64	170	60	26	Left	Infarct ion	44	18

C. SCHEME OF FEEDBACK

The purpose of this study is to induce active movement during specific phases of the arm swing cycle divided according to the side (left or right) and direction (forward or backward) of movement. These arm swing phases can be accurately explained using the out-of-phase coordination with the lower limb (e.g. Left leg swing phase → Right arm moving forward & Left arm moving backward) [31]. However, since the arm swing can simply be assumed as the motion of a pendulum [15], [31], to help understanding, the arm swing phases in this study are divided based on the heel strike of the lower limb, as shown in Fig. 4 (a). The trial conditions used in the healthy subject study are detailed in section III A.



(a) Relationship of the trial conditions with the subject's gait phases (left arm forward & right arm backward movements coincide with the right step (DS1 & Right swing phase) and right arm forward & left arm backward movements coincide with the left step (DS2 & Left swing phase))



(b) Operational flow of the system control software.

FIGURE 4. Relationship of upper limb movement phases with lower limb movement and operational flow of the system control software.

During each trial, the arm swing angle was measured in real time by the developed system and when the desired angle was reached, the vibration of the arm in question was turned on to signal to the subject that they had reached their goal (see fig. 4 (b)). This intuitive feedback scheme was adopted because people generally have a preference for intuitive feedback schemes [32].

III. HEALTHY SUBJECTS STUDY

A. PROTOCOL

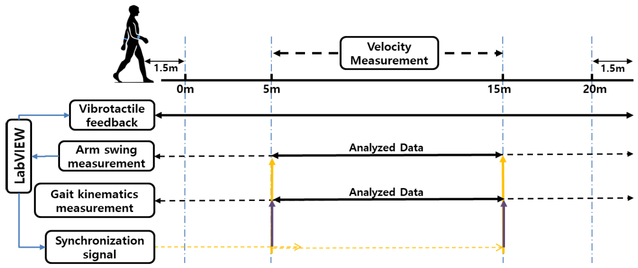
The healthy subjects were asked to perform two gait trials under each of the conditions given below. A break of 2 minutes was given between trials. The subjects were asked to take seated rest after every fifth walking bout or if they felt the need to do so. The following trial conditions were used in this study:

1. **NW**: Normal Walking (No Vibration Feedback)
2. **LF**: Left Forward
3. **RF**: Right Forward
4. **RB**: Right Backward

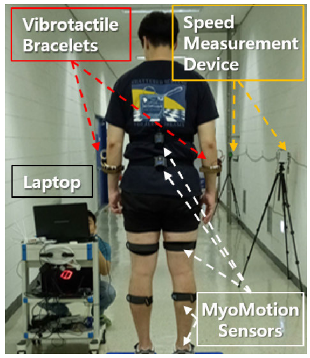
5. **LB**: Left Backward
6. **LFRB**: Left Forward & Right Backward
7. **RFLB**: Right Forward & Left Backward
8. **BFBB**: Both Forward and Both Backward

This study was approved by the Bioethics Review Committee of Gwangju Institute of Science and Technology, South Korea (20201008-HR-56-03-04) and all the participants provided informed consent prior to the start of trials. As shown in fig. 4 (b), before the start of each walking bout, the subjects were asked to stand straight with their arms hanging freely by their sides and the IMU sensors of both the vibrotactile bracelets were calibrated to set the zero position for measuring the arm swing. Arm movement towards the front of this position is considered as positive and vice versa. Once the setup was complete, the subjects were asked to walk on a straight path at their preferred walking speed while trying to achieve the particular trial goal. The target arm swing angle during all trials was a 100% increase in the baseline swing value (obtained from the normal walking (NW) trial) of the particular arm and swing direction that was different for each participant. For example, in the left forward (LF) trial condition the subject had to increase the forward swing of their left arm to reach the goal representing a 100% increase in the left arm forward swing recorded during their NW trial. Subjects were asked to swing the non-targeted arm naturally as they do during walking so that it may adapt to the changed kinematics. This is to observe the effects of arm swing modification on the movements of the untargeted arm and direction, and to maintain consistency between the healthy and stroke subject protocols as we cannot restrict the non-targeted arm swing of the stroke patient.

As shown in fig. 5 (a), the subjects walked in a straight line for 20m at their preferred pace during each walking bout. They started walking from a point that was set 1.5m prior to the starting line. This allowed them to reach their preferred walking speed as they entered the 20m walking path. Similarly, to keep the slow down period at the end of the walking bout out of the 20m path, they were asked to stop at a point set 1.5m beyond the end of the 20m path. In order to prevent cognitive dissonance, the arm was moved in place before departure so that the subject could feel a few vibrations before walking. In addition, the subjects were asked to gradually increase their arm swing while passing through the initial acceleration section. Thus, all the subjects were able to meet the trial requirements in a comfortable way. If deemed necessary by the researchers, the subject was allowed to practice walking a few steps under the trial conditions before the experiment. The data obtained from the mid 10m of the walking path were used for analysis. The walking speed was measured using a 'walkthrough gate' type gait speed measurement system (SR500, SeedTech, Korea). The system consists of two optical sensors 'gates'. The first gate was placed at the 5m mark and the second was placed at the 15m mark to measure the exact speed during the analyzed 10m section. IMU based motion capture equipment (MyoMotion, Noraxon, USA) worn by the subjects on their



(a) The experiment procedure



(b) The experiment environment

FIGURE 5. The experiment procedure and environment.

feet, calves, thighs, pelvis and lower thorax tracked their gait kinematics and walking balance. The data recorded by this system was synchronized with the arm swing data recorded by the feedback system software using the MyoSync device (Noraxon, USA). The complete experimental setup is shown in Figure 5 (b). Figure 6 shows the device data of the left arm swing during one complete trial of NW and LF. The two data are synchronized according to the first pulse of the synchronization signal (Sync (NW, LF)).

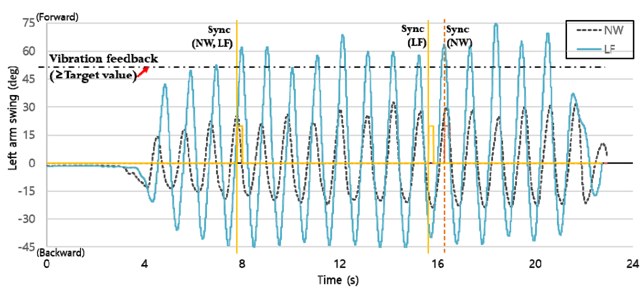


FIGURE 6. The left arm swing data of a representative subject gathered during the NW and LF trials using the developed system (The yellow colored square pulses are the synchronization signals).

B. DATA ANALYSIS

The gait speed of the healthy subjects was recorded using the gait speed measurement device while the arm swing angles were measured using the developed vibrotactile bracelets and all the data were logged on the laptop. Stride length and step length were reported by the MyoResearch (MR3 3.14, Noraxon, USA) software based on data recorded by the motion capture IMUs worn on the lower limbs, pelvis and lower thorax. In addition, by extracting pelvic tilt value from the MR3 report, RMS values of Mediolateral (ML) and Anteroposterior (AP) tilts were calculated to determine the balance conditions during the various gait trials. The symmetry ratio for the healthy subjects ($SR_{Healthy}$) calculated based on the step length [33] using equation (1) was used to determine the direction of increase (right or left) of the step length caused by the increased arm swing.

$$SR_{Healthy} = \frac{StepLength_{Left}}{StepLength_{Right}} \tag{1}$$

The symmetry ratio for the stroke patient (SR_{Stroke}) was calculated using the following equation.

$$SR_{Stroke} = \frac{StepLength_{Paretic}}{StepLength_{Non-paretic}} \tag{2}$$

According to [33], it is recommended to always use the smaller value as the denominator in order to avoid any statistical analysis errors (for example, 1.2 over 1 and vice versa represent different values of 1.2 and 0.83). However, since we are interested in knowing not just the symmetry ratio but also the side (left or right) of the step length increase, equations (1) and (2) are used as they are under all conditions presented in this paper. In addition, as shown in the example, the statistical error acts as a factor that makes the significant difference smaller rather than larger.

For post-experimental data analysis, a one-way repeated measures analysis of variance (RMANOVA) was performed to study the effects, at preferred walking speed, of the various changes in arm swing, which had seven conditions (LF, RF, RB, LB, LFRB, RFLB, BFBB) on gait speed, stride length, symmetry ratio, and RMS values of ML and AP tilts. Mauchly's test of Sphericity was used to confirm the validity of the RMANOVA results. Post hoc tests were conducted using the Bonferroni correction method and Greenhouse-Geisser corrections were used where Mauchly's test of sphericity was violated. Partial eta squared was calculated as a measure of the effect size for one-way RMANOVA. All statistical analysis was carried out using SPSS V26.0 (IBM Corp., USA).

C. RESULTS

1) ANGLE OF ARM SWING

Figure 7 shows the mean and standard deviation values of the Angle of Arm Swing (AAS) of healthy subjects in each trial. The above zero values represent movement in the forward direction while the below zero values represent movement in the backward direction. Fig 7 (b) shows the rate of increase of the AAS (%). Here, the value of AAS during NW is taken as 100% (baseline value) and shown as the black dotted line. So, the target values for the other trials are 200% (100% increase in baseline value), which is represented by the green dotted line. As shown in Fig. 7 (b), targeted directional arm swing increase was successfully accomplished during each protocol. However, a relatively smaller amount

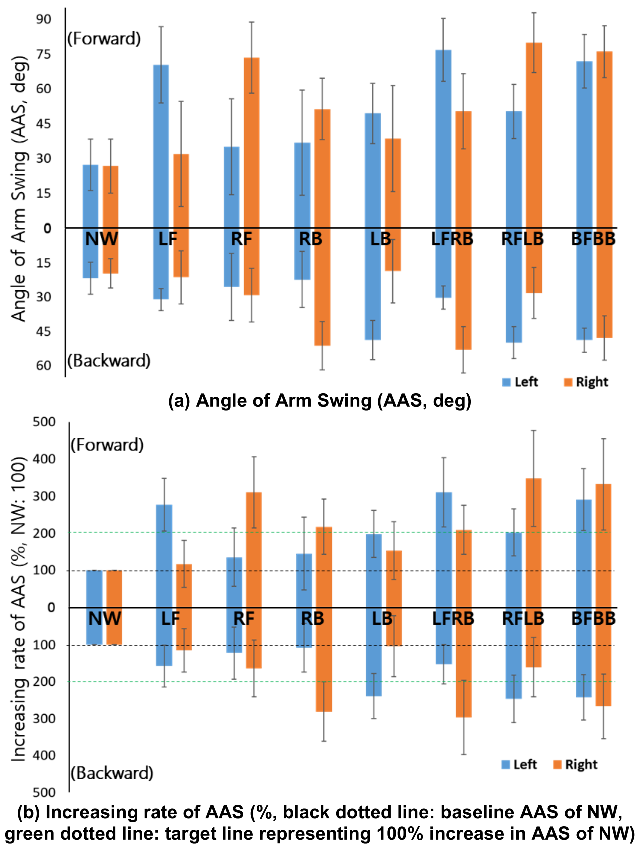


FIGURE 7. Mean and SD of the range of arm swing of healthy subjects during various gait trials (above zero: forward, below zero: backward). The error bars represent the standard deviation.

of increase in swing of the non-targeted arm is observed to accompany the increase in the swing of the targeted arm. This effect is more pronounced during protocols targeting backward movement. In Table 3, the swing angle values of the targeted arm and direction during each protocol are written in bold typeface and are underlined.

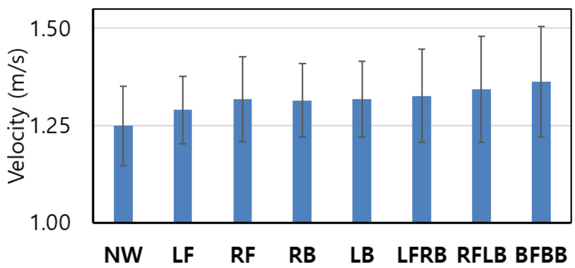
2) GAIT PARAMETERS

Fig. 8 (a) shows the gait velocity (m/s) while fig. 8 (b) shows the stride length (cm) of the healthy subjects during each protocol. The mean gait velocity increased with increasing participation of arms (LF, RF, RB, LB vs LFRB, RFLB vs BFBB) in the targeted swing, but the RMANOVA revealed no statistically significant differences in velocity between the different trial conditions. The stride length also increased with the same trend. However, in this case the RMANOVA revealed significant differences in stride length between several trial conditions ($F(2.49, 22.37) = 13.01$, $p < .001$, $\eta_p^2 = .59$): NW and LF ($p < .05$), NW and RF ($p < .01$), NW and RB ($p < .005$), NW and LB ($p < .005$), NW and LFRB ($p < .01$), NW and RFLB ($p < .05$), and NW and BFBB ($p < .005$).

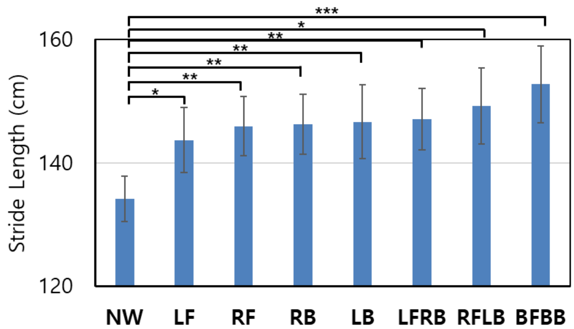
The results of the one-way repeated measures ANOVA of Symmetry ratio (SR) of healthy subjects are presented in Fig. 9. SR was calculated based on the step length

TABLE 3. AAS and step length of healthy subjects (AAS value of targeted side and direction of arm(s) is highlighted).

Protocols		NW		LF		RF		RB		LB		LFRB		RFLB		BFBB	
Side		Left	Right	Left	Right	Left	Right	Left	Right	Left	Right	Left	Right	Left	Right	Left	Right
Angle of arm swing (deg)	Forward	27.3 ±11.1	26.7 ±11.7	**70.4** ±16.4	31.9 ±22.6	35.0 ±20.6	**73.5** ±15.3	36.8 ±22.7	51.3 ±13.3	49.4 ±13.0	38.6 ±22.9	**76.7** ±13.5	50.3 ±16.3	50.3 ±11.7	**79.9** ±12.9	**71.8** ±11.5	**76.1** ±11.2
	Backward	21.7 ±7.0	19.6 ±6.4	31.0 ±4.8	21.2 ±11.6	25.6 ±14.6	29.2 ±11.8	22.2 ±12.4	**51.2** ±10.6	**48.8** ±8.6	18.6 ±13.9	30.2 ±5.0	**53.0** ±10.2	**49.9** ±7.0	28.2 ±11.2	**48.9** ±5.3	**47.9** ±9.7
Step length (cm)		67.2 ±3.7	67.0 ±3.9	70.8 ±5.3	72.8 ±5.6	73.7 ±4.8	72.2 ±4.0	72.9 ±4.9	73.4 ±5.3	73.4 ±6.0	73.2 ±4.3	72.2 ±5.0	74.9 ±6.0	75.9 ±6.2	73.4 ±5.7	76.3 ±6.2	76.5 ±5.5



(a) Gait velocity (m/s)



(b) Stride length (cm)

FIGURE 8. Mean and SD (standard deviation) values of the gait speed and stride length results of healthy subjects during various gait trials. Statistically significant differences are marked based on Post-hoc pairwise comparisons ($*$ = P-value < 0.05, $**$ = P-value < 0.01, $***$ = P-value < 0.001)). The error bars represent the standard deviation.

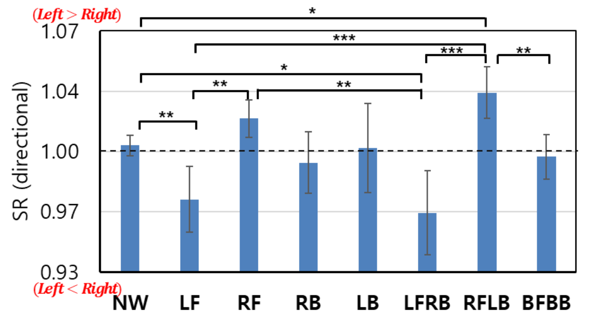
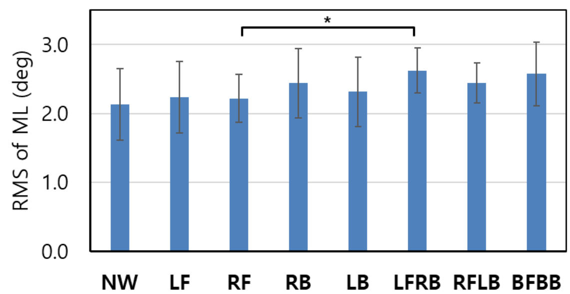
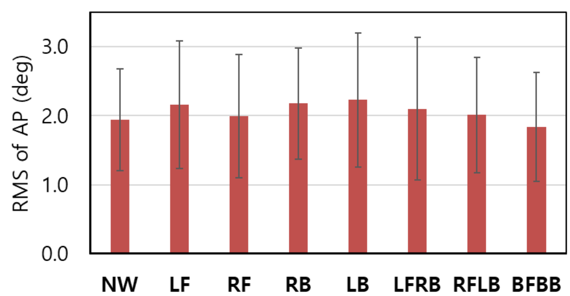


FIGURE 9. Mean and SD values of the symmetry ratio (SR) of healthy subjects during various gait trials calculated using eq. (1) (SR > 1: Left > Right, SR < 1: Left < Right). Statistically significant differences are marked based on Post-hoc pairwise comparisons ($*$: p < 0.05, $**$: p < 0.01, $***$: p < 0.001)). The error bars represent the standard deviation.



(a) RMS of ML tilt at the pelvis



(b) RMS of AP tilt at the pelvis

FIGURE 10. Mean and SD values of the RMS of ML & AP tilts of healthy subjects. Statistically significant differences are marked based on the result of Post-hoc pairwise comparisons ($*$: p < 0.05). The error bars represent the standard deviation.

using equation (1). Statistical analysis revealed significant differences in SR between several trial conditions ($F(2.71,24.39) = 10.19$, $p < .001$, $\eta_p^2 = .53$): NW and LF ($p < .005$), NW and LFRB ($p < .05$), NW and RFLB ($p < .05$), LF and RF ($p < .005$), LF and RFLB ($p < .001$), RF and LFRB ($p < .005$), LFRB and RFLB ($p < .001$), and RFLB and BFBB ($p < .005$).

3) ML & AP TILTS

The results of the one-way RMANOVA of RMS of ML & AP tilts of healthy subjects are presented in Figure 10. Analysis of RMS of ML and AP tilts revealed no significant differences between different trial conditions except for a significant difference between ML tilts recorded during RF and LFRB trials ($F(2.81,25.29) = 4.76$, $p = .05$, $\eta_p^2 = .35$ for RMANOVA and $p < .05$ for post-hoc pairwise comparison).

D. DISCUSSION

The objectives of this section of the presented research are to study how gait and balance parameters change when the arm swing is modified using our developed vibrotactile biofeedback bracelets, and to determine the possibility of applying this methodology for the gait rehabilitation training of stroke patients. Therefore, we designed a protocol that includes all the combinations of arm swing increase with respect to the different phases of arm swing.

From Fig. 8 (a) and (b), it can be deduced that increase of arm swing angles has a more profound influence on the stride length. Stride length (cm) and gait velocity (m/s) both increased with the increased arm swing, but the stride length showed significant differences compared to NW while the walking speed did not show any significant change. This may be because the amplitude of the increased arm swing has the effect of increasing the amplitude of the stride. This result is supported by previous research that reported a postural compensation phenomenon in the reverse direction in the sense that deliberate decrease or increase of the arm swing amplitude results in a decrease or increase of stride length [18]. Since active muscle control of arm swing during normal walking is required to obtain an out-of-phase coordination with the legs [31], [34], it can be considered that the increased muscle activity required to increase the swing also contributes to the increase in stride length. Eke-Okorol *et al.* found that inducing a swing that makes both arms reach 'full excursion' (more than 90 deg) increases the stride length, but this was accompanied by a decrease in the walking speed [18]. In the current study, the average value of gait velocity showed an increase between NW (1.25 m/s) and BFBB (1.36 m/s, L: 72°, R: 76°) trials. This phenomenon needs to be studied further to examine the optimal relationship between arm swing modification and stride length and gait velocity.

As shown in fig. 9, significant differences in SR were present between NW and LF, LFRB, RFLB conditions. However, for the RB and LB trials, left and right direction differences were apparent in the mean values, but the results were not significantly different from each other. This may be due to the phenomenon that when the backward arm swing was increased, the forward arm swing also increased by around 100% (fig. 7 (b), RB: (R) $+115\% \sim -181\%$, LB: (L) $+98\% \sim -139\%$). Thus, it is apparent from the current results that protocols containing variation of the forward directed arm swing has greater efficacy in modifying gait symmetry as protocols with only backward directed arm swing variation did not have a significant effect on gait symmetry. In the LF, RF, LFRB, and RFLB trials, the symmetry ratio clearly differed between left and right arm conditions, resulting in significant differences between results (LF vs RF, LF vs RFLB, RF vs LFRB and LFRB vs RFLB). In a previous study, it was found that there is a correlation between diagonal upper/lower limb trajectories, which increased with the increase in the right belt speed of a split belt treadmill [13]. The authors mentioned that this relationship is most likely due to the body's neural mechanisms in which the central pattern generators (CPGs) of the upper and lower limbs regulate full-body movement and maintain the rhythmic locomotor pattern [13], [15]. The current study revealed the existence of a relationship between the upper and lower limbs whereby the modification of the arm swing causes a change in the movement of the corresponding lower limb. This relationship is similar but opposite in direction to the one reported in the previous studies mentioned above. However, as shown in fig. 8, it is interesting to note that the backward directed modifications (RB and LB, ($p < .005$)), result in more significant stride length variations than LFRB and RFLB conditions ($p < .01$, $p < .05$), in which the forward swing of one arm and the backward swing of the other arm were increased together. According to Patterson *et. al.,* there is a relationship between the step length symmetry ratio and the propulsive force during gait [33]. Thus, in the future, gait analysis studies with ground reaction force or electromyographic sensors are required to further explore the effects of the directional (forward and backward) relationship between arm swing variation and lower limb movement during gait.

In this study, during NW trials, the left-arm swing was observed to be slightly greater than the right-arm swing. This is consistent with the observations of previous researchers who have reported that the arm swing is not completely symmetrical [34]–[36]. Killeen, Tim, *et al.* reported that the normal arm swing is not related to the dominant arm [36], however in the current study it can be observed that more active arm swing may have a relationship with the dominant arm. As shown in Fig. 7, while comparing the left arm and right arm symmetric protocols (LF vs RF, RB vs LB, LFRB vs RFLB, BFBB), the range of swing of the right (dominant) arm is slightly larger than that of the left arm. Since no significant differences were found between the RMS of ML and AP tilts during increased arm swing trials and those during NW trials, it is determined that the tested protocols have no adverse effect on gait balance. Thus, they can be safely used to induce gait modifications through arm swing variations.

IV. STROKE PATIENT PILOT STUDY

A. PROTOCOL

The pilot study with a stroke subject was conducted following the principles of the Declaration of Helsinki under the supervision of a rehabilitation medicine doctor at the Department of Rehabilitation Medicine at Gyeongsang National University Hospital, South Korea. The subject gave written informed consent prior to data collection.

The test environment for the stroke patient was similar to that used for the healthy subjects, except for the reduced lengths of both the acceleration and deceleration sections (2.5m instead of 5m). This modification was done to minimize patient fatigue, which can accumulate over several trials. In each trial, a rehabilitation doctor followed 1 to 2 steps behind the subject to ensure their safety in case of any untoward incident.

The overall protocol for stroke subject testing was the same as that used for the healthy subjects. However, the number of trial conditions were reduced as only the best performing trial conditions from the healthy trials were used in the stroke trial (this selection is further explained in the discussion section). The trial conditions used are detailed below.

1. **NW**: Normal Walking (No Vibration Feedback)
2. **PF**: Paretic side arm Forward
3. **NB**: Non-paretic side arm Backward
4. **NF**: Non-paretic side arm forward
5. **PB**: Paretic side arm Backward
6. **PFNB**: Paretic side arm Forward & Non-paretic side arm Backward
7. **BFBB**: Both Forward and Both Backward

B. RESULTS

1) ANGLE OF ARM SWING

Fig. 11 (a) shows the mean and SD values of AAS of the stroke subject during each gait trial, while the rate of increase of AAS is shown in Fig. 11 (b). The subject (whose left side was paralyzed) could successfully accomplish the required arm swing modifications during all trials except for the backward arm swing during PFNB and PFBB trials.

2) GAIT PARAMETERS

Velocity (m/s) and stride length (cm) results of the stroke subject are shown in Fig. 12 (a) and (b), respectively. Both these quantities showed an increase as compared to NW, and exhibited trends similar to each other. The mean and SD values of SR of the stroke subject are shown in Fig. 13.

3) ML & AP TILTS

The mean and SD values of the RMS of ML and AP tilts of the stroke subject are shown in Fig. 14.

C. DISUCUSSION

The healthy subject results (section III C) showed that the largest value of RMS of ML tilt occurred during the LFRB trials, and a diagonal relationship between the upper and lower limbs was apparent from the symmetry ratio with increased stride length. Therefore, only one protocol involving both arms that could lengthen the step length of the shorter step of the stroke subject (PFNB) was included in the stroke study. Furthermore, the protocol that increased the swing of both arms in both directions (BFBB) was also included in the stroke study as it produced the greatest increases in gait speed and stride length. Inclusion of this condition is also supported by the findings of A. Hill and J. Nantel, who showed that active increase of arm swing (more than 90°) of both arms increased the local trunk stability of young healthy subjects during asymmetric walking on a split-belt treadmill (left/right, 5:4 speed ratio) [37]. The number of multi-arm protocols was reduced in consideration the stroke survivor's low cognitive ability that may lead to a decrease in speed [36] and may cause balance problems [38].

In individuals with hemiplegia after stroke, the paretic arm usually swings with decreased amplitude [15], which is also evident from the results of the current study where the affected side showed a much smaller swing angle. However, the stroke subject was able to meet the requirements of all

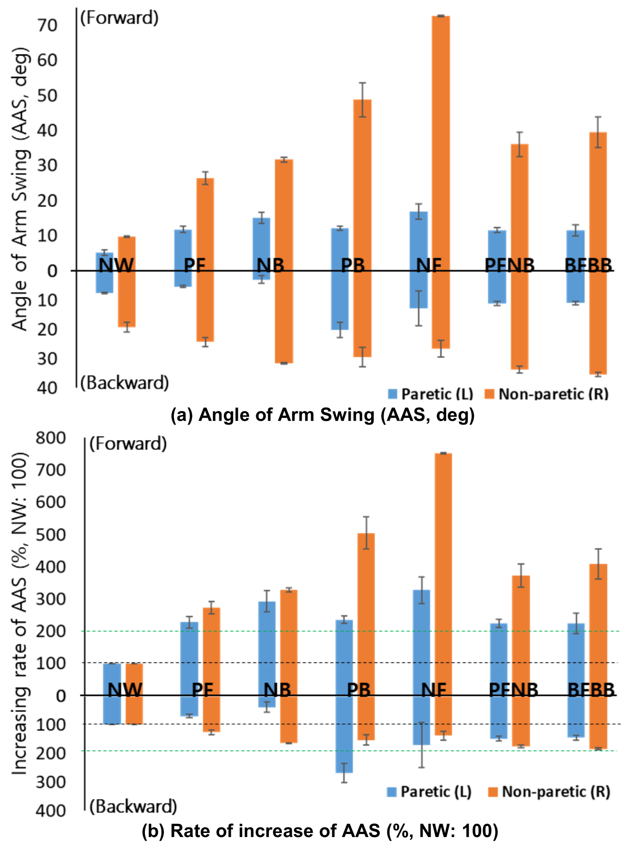


FIGURE 11. Mean and SD of the angle of arm swing results of the stroke subject during various gait trials. The error bars represent the standard deviation.

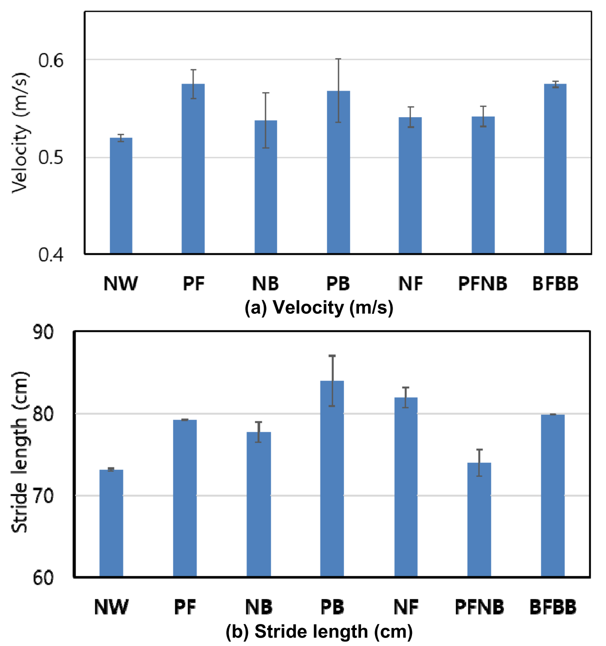


FIGURE 12. Mean and SD values of the gait velocity and stride length results of the stroke subject during various gait trials. The error bars represent the standard deviation.

TABLE 4. Specific values of AAS and step length of the stroke subject (AAS value of targeted side and direction of arm(s) is highlighted in each protocol).

Protocols		NW		PF		NB		PB		NF		PFNB		BFBB	
Side		Left	Right	Left	Right	Left	Right	Left	Right	Left	Right	Left	Right	Left	Right
Angle of arm swing (deg)	Forward	5.1 ±0.7	9.7 ±0.3	**11.7** ±0.9	26.4 ±1.8	15.1 ±1.7	31.7 ±0.7	12.1 ±0.6	48.8 ±4.8	16.8 ±2.1	**72.7** ±0.3	**11.6** ±0.7	36.0 ±3.4	**11.5** ±1.6	**39.5** ±4.4
	Backward	7.5 ±0.2	19.3 ±1.6	5.3 ±0.4	24.3 ±1.6	2.9 ±1.3	**31.6** ±0.2	**20.2** ±2.6	29.5 ±3.4	12.9 ±6.0	26.6 ±2.8	11.2 ±0.7	**33.8** ±1.1	**11.0** ±0.6	**35.4** ±0.8
Step length (cm)		39.3 ±0.3	33.9 ±0.4	41.2 ±0.7	37.9 ±0.5	40.5 ±0.1	37.2 ±1.4	42.7 ±1.2	41.4 ±2.0	39.9 ±0.7	42.1 ±0.4	38.2 ±1.1	35.9 ±0.4	40.6 ±0.3	39.2 ±0.3

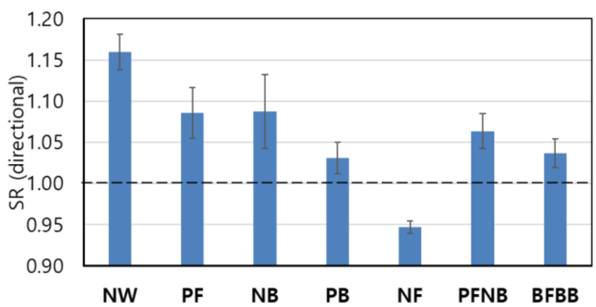


FIGURE 13. Mean and SD values of the symmetry ratio (SR) of the stroke subject (Paretic side: Left) during various gait trials calculated using eq. (2) (SR > 1: Left > Right, SR < 1: Left < Right). The error bars represent the standard deviation.

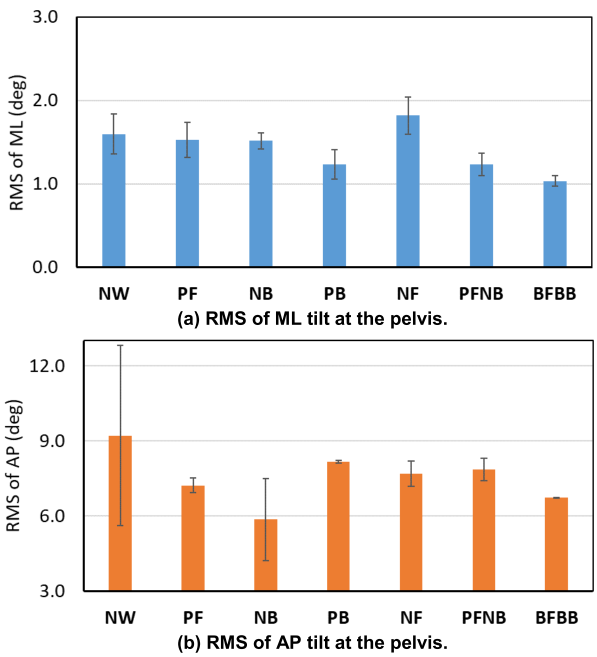


FIGURE 14. Mean and SD values of the RMS of ML & AP tilts of the stroke subject. The error bars represent the standard deviation.

the protocols, but it can be seen from the results that accurate arm swing control is difficult in case of a large arm swing amplitude, as in the case of the NF trial. In addition, it was observed that increase in the non-paretic arm swing caused an increase in the paretic arm swing, which is in agreement with the previously reported observation that the paretic side is dependent on the non-paretic side [39].

Gait velocity (m/s) and stride length (cm) of the stroke subject during arm swing modification trials were increased compared to NW trials. Ustinova *et al.* showed that increasing the amplitude of arm swing and out-of-phase synchronization of both arms through instruction improved the step length of patients with Traumatic Brain Injury (TBI), but decreased their gait velocity [40]. It suggests that for stroke patients who have low cognition, trunk rotation due to arm swing is more important than synchronization. Yang *et al.* showed that an increase in gait speed was observed when healthy subjects walked with additional unilateral arm and waist weights, rather than bilateral arm weights [26]. Thus, since stroke patients are expected to require greater trunk rotation to enhance the arm swing [19], arm swing variations may have a greater effect on gait speed changes.

When looking at the symmetry ratio (SR), all the arm swing modification protocols resulted in improvements compared to the NW trials. In particular, it was observed that the non-paretic arm forward, paretic arm backward and BFBB protocols, which were related to the paretic step length of the stroke subject, showed greater improvements in SR. However, the findings of this pilot trial need further evaluation through additional experiments with stroke survivors. It has been shown that neural deficits of the paretic arm are compensated for by the influences from the unaffected side [39]. Furthermore, Stephenson *et al.* showed that asymmetric muscle activity in the lower limbs was activated during treadmill waking with arm movement [22]. Thus, in the future, the exploration of how the improvement of symmetry due to enhanced arm swing affects the muscle activity of the lower extremities is required. Stroke patients usually suffer from hemiplegia that makes it difficult for them to modify the movement of their paretic leg by themselves. A cross coupled relationship between arm movement and leg behavior may be beneficial for stroke rehabilitation as it can allow improvement of gait parameters by affecting the spinal CPGs through the modification of arm movements [15].

In healthy subjects, the protocols requiring greater arm participation, such as LFRB and RFLB, had a greater influence on the gait parameters. However, in case of PFNB in the stroke subject trials, there was only a small improvement in speed, stride length, and symmetry ratio. In particular, the absolute difference of swing angles between PFNB (L: $+11.6\pm0.7^\circ \sim -11.2\pm0.7^\circ$, R: $+36.0\pm3.4^\circ \sim -33.8\pm1.1^\circ$) and BFBB (L: $+11.5\pm1.6^\circ \sim -11.0\pm0.6^\circ$, R: $+39.5\pm4.4^\circ \sim -35.4\pm0.8^\circ$) is quite small (Table 4), but the differences in the resultant walking speed and stride length are quite large. This is believed to be because greater cognitive ability is required to accomplish the PFNB protocol. Therefore, in future patient experiments, only one arm and both arm (BFBB) protocols are considered to be sufficient.

V. CONCLUSION

The aims of this research were to study how gait parameters are affected when the arm swing is modified using biofeedback delivered by the developed vibrotactile bracelets, and to check the usability of this methodology for gait rehabilitation training of stroke patients. Therefore, we designed a protocol that included all combinations of arm swing modifications in terms of arms and swing phases involved and analyzed the results of experiments carried out with ten healthy subjects and one stroke subject.

From the healthy subject experiments, we found that increasing the arm swing increased the stride length and gait velocity. However, while the increase in stride length was statistically significant, the gait velocity did not show a significant change. Furthermore, we found that a relationship exists between the upper and lower limbs where variations in the arm swing amplitude affect the movement of the corresponding leg, thus affecting the gait symmetry ratio. It was also observed that there were no significant differences in the RMS of ML & AP tilts as compared to those during NW trials.

In the pilot test with a stroke subject, the subject was able to successfully perform almost all the arm swing modifications and the results showed an increase in the stride length and velocity. The arm swing modifications also affected the symmetry ratio but further studies with stroke subjects are required to fully evaluate the effects of these modifications.

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