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Development of Abnormal Gait Detection and Vibratory Stimulation System on Lower Limbs to Improve Gait Stability

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Abstract The purpose of this study is to develop an abnormal gait detection algorithm and a vibratory stimulation system on a lower limb to improve gait stability and prevent falls. The system consists of a gait measurement module, an abnormal gait detection module, and a vibratory stimulation module. The gait measurement module measures the vertical acceleration of the ankle during walking using an accelerometer. The measured acceleration values are sent to a portable microcontroller, which controls vibratory stimulations to the ankles based on an algorithm that detects the peak acceleration values. If the acceleration peaks are found to occur irregularly, the abnormal gait detection algorithm activates the vibratory stimulation module. To determine the effect of vibratory stimulations under dynamic condition, this study investigated the contribution of ankle muscle proprioception on the control of dynamic stability and lower limb kinematics while walking using vibratory stimulation to alter the muscle spindle output of individuals' left lower limb. Vibrators were attached to the left ankle joint (tibialis anterior, triceps surae). Participants were required to walk along a travel path and step over an obstacle placed in their way. There were four task conditions; an obstacle (10%, 20%, and 30% of the participants' height) was positioned at the midpoint of the walkway, or the participants' walking path remained clear. For each obstacle condition, participants experienced either no vibration, or vibration of the tibialis anterior muscle and the triceps surae muscle of the left lower limb. Vibration began upon detection of an abnormal gait and continued for one second. Vibrating the ankle muscles of the left lower limb while stepping over an obstacle resulted in significant changes in COM behavior on both the anterior/posterior (A/P) and medial/lateral (M/L) planes. The results provide strong evidence that the primary

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endings of the ankle muscle spindles play a significant role in the control of posture and balance during the swing phase of locomotion by providing information on the movement of the body's COM with respect to the support foot.

Keywords Postural adjustment · Gait stability · Abnormal gait detection · Muscle vibration

Introduction

In order to control human posture, the body needs a process of integrating information input from the visual, somatic, and vestibular senses in the central nervous system to induce reflexive control of the body [1]. Senses necessary for posture control are weakened with aging, and provide weakened sensory information or inappropriate feedback to the system. Such a loss of physical functions resulting from aging degrades the ability to walk and increases the risk of fall [2]. It was reported in Korea that 49.3% of elderly people aged over 60 experienced a fall [3], and 25.2% of elders aged over 65 [4]. The aged population shows a much higher death rate resulting from a fall than younger populations, and suffer greater injury. When elderly people are wounded by a fall, not only their personal life, but also the psychological, physical, and economic burden of medical expenses cause social problems [5]. In order to prevent fractures resulting from falls, the Fall Prevention Pad and Safe Hip were developed [6]. Such pads have air or a buffer inside to absorb shock around the hips when a fall occurs. These devices can prevent fractures, but are uncomfortable and unpleasant to wear at high temperature. Research is being performed to prevent wounds caused by falls, but there are not many studies for preventing falls actively by securing posture stability using sensory information.

The method of applying vibration to muscles to enhance posture stability is being used widely in research for improving somatic sensation. Inglis et al. [7] proved that, during the voluntary motion of the upper limbs, the vibration of an antagonist plays an important role in recognizing motion and body orientation. In addition, Sorensen et al. [8] proved that vibratory stimulation applied to the ankle muscle spindles provides the central nervous system with information regarding transfer of the body's gravity center in the lower limb and, by doing so, plays an important role in controlling posture and balance during locomotion.

The present study implemented an abnormal gait detection algorithm for improving gait stability, and developed a vibratory stimulation system that secures gait stability through abnormal gait detection and vibratory stimulation to prevent falls. In addition, we implemented an algorithm evaluation program for evaluating the effectiveness of the abnormal gait detection algorithm.
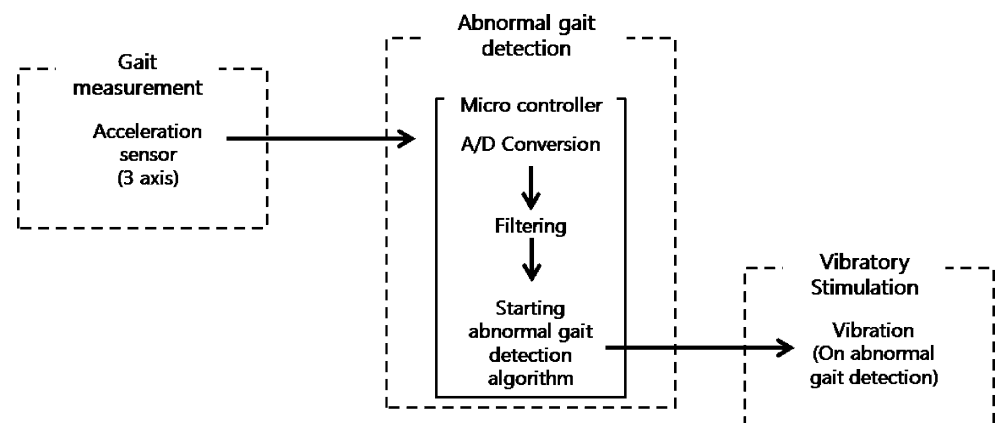
Abnormal gait detection and vibratory stimulation system for improving gait stability

The present system is composed of a gait measurement module, an abnormal gait detection module, and a vibratory stimulation module. Figure 1 shows the schematic diagram of the system. It has an accelerometer for measuring gaits, a microcontroller for detecting abnormal gaits, and a vibration motor for applying vibratory stimulation. During a gait, the gait measurement module measures acceleration using the accelerometer and sends the measurements to the microcontroller. If the microcontroller programmed with the abnormal gait detection algorithm detects an abnormal gait from the peak acceleration values, the vibratory stimulation module generates vibratory stimulation.

Gait measurement module

The gait measurement module, which measures the vertical acceleration of the ankle joint during a gait, was implemented as shown in Fig. 2a. A MMA7260Q (Freescale Semiconductor, Inc., USA) accelerometer was used, which measures three-axis (X, Y, and Z) acceleration

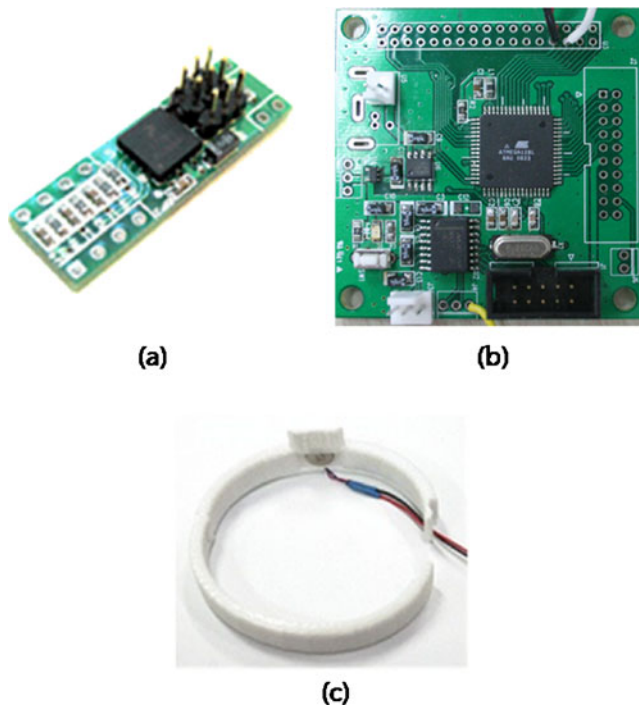**Fig. 1** The overall system

Fig. 2 The overall system: **a** Gait measurement module, **b** Abnormal gait detection module, and **c** Vibratory stimulation module

and can be easily connected to other circuits or micro-controllers by installing a peripheral circuit for operating the accelerometer in PCB. The accelerometer was implemented to be operable simply by applying power without additional parts. The acceleration measuring range is 1.5 g/2 g/4 g/6 g; if the range is narrowed, resolution increases, and if the range is widened, resolution decreases.

The gait measurement module was a small, low-power type (500 μA), and was embedded with a low-pass filter for measuring acceleration. In addition, the acceleration measurement range was set to 0~4 g, and the resultant resolution was 300 mV/g.

Abnormal gait detection module

The abnormal gait detection module, which detects abnormal gaits from the peak acceleration values measured by the accelerometer, was implemented as shown in Fig. 2b. The microcontroller used to detect abnormal gaits was an Atemga128L (8-bit microcontroller, 128 kbytes In-System Programmable Flash, 4 Kbytes EEPROM), which is a high-performance low-power 8-bit microcontroller. The module receives the input of vertical acceleration from the accelerometer through the ADC channel. Then, the input goes through A/D conversion and filtering, and abnormal gaits are detected by the abnormal gait detection algorithm explained below.

Vibratory stimulation module

The vibratory stimulation system is used to prevent falls by applying vibratory stimulation during a gait, thereby inducing tension in muscle. The module used a flat vibration motor (JHV-10A1, JAHWA Electronics Co.) 3.4 mm in height, 10 mm in diameter, and 10 g in weight. Figure 2c shows the vibratory stimulation system mounted with a vibration motor.
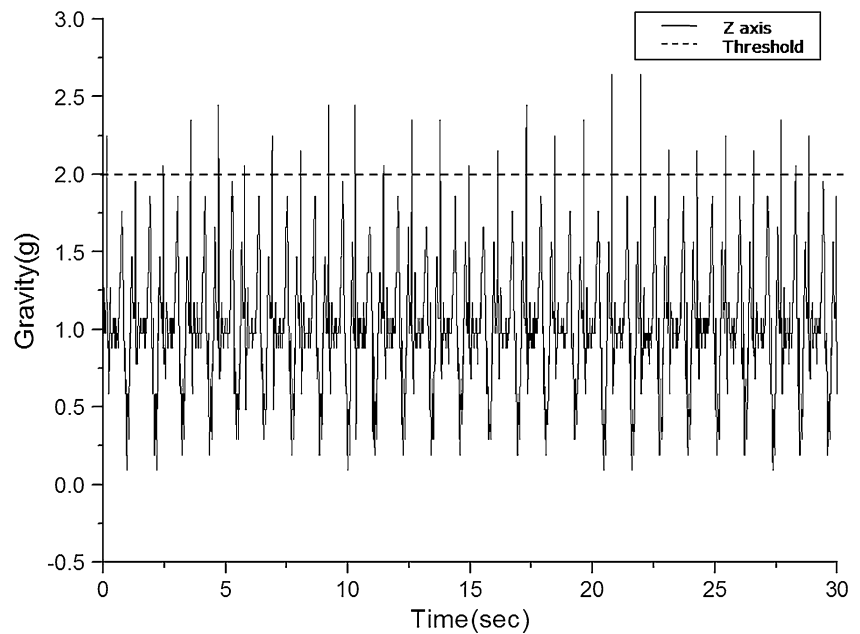
Development of abnormal gait detection algorithm

The abnormal gait detection algorithm monitors the peak values of acceleration during a gait when the foot touches the ground, and detects an abnormal gait based on the variation of time interval between peak values. If the variation of time interval is large, it is considered an abnormal gait and vibratory stimulation is applied, while if the variation is not large, a normal gait is assumed and vibratory stimulation is not applied.

Acceleration used to develop an algorithm in this study is vertical acceleration measured in the ankle joint. Figure 3 shows the results of an experiment with ten people, in which the peak values of acceleration were measured using an accelerometer when the foot touched the ground during a gait. According to the results, the peak values of acceleration were over 2 g (gravity) when the foot touches the ground, and less than 2 g in other cases.

The flow of the abnormal gait detection algorithm is shown in Fig. 4. Before the algorithm is started, all parameters are initialized to be 0. In order to distinguish the moment when the sole touches the ground during a gait, the algorithm was set to run when the peak value of acceleration was greater than 2 g. In the state that AdcTime (timer) increases by 1 in every 10 ms, acceleration values generated from the accelerometer go through A/D conversion and are saved in AccAdc (current acceleration). The acceleration value is updated every 10 ms, and if it is less than 2 g, the algorithm ends without executing the remaining part of the algorithm. If acceleration is 2 g or higher, AdcTime (time at the moment) is saved in PeakTimeCur (current peak time). Then, the difference between the current peak time (PeakTimeCur) and previous peak time (PeakTimePrev) is saved in PeakTimeDiffCur (the current peak time difference). Finally, the current peak time is transferred to the previous peak time. If the current and the previous peak time difference (PeakTimeDiffPrev) are smaller than parameter value δ, the gate is considered normal and vibratory stimulation is not generated. However, if the difference is δ or larger, an abnormal gait is detected and vibratory stimulation is generated. The algorithm above was implemented into a firmware program using AVREdit (AvrGcc).

Fig. 3 Vertical acceleration of the ankle joint during normal gait

Evaluation of the abnormal gait detection algorithm and results

In order to evaluate the abnormal gait detection algorithm, we implemented an algorithm evaluation program in Visual C++ as shown in Fig. 5. The output parameters in the abnormal gait detection algorithm are transmitted to the algorithm evaluation program through serial communication. The algorithm evaluation program displays the transmitted parameters on a window, and acceleration is represented in a graph. In the graph, the straight line indicates 2 g. When an acceleration value exceeds 2 g, the current peak time is displayed and, consequently, the previous peak time difference and the current peak time difference. In addition, when the two values above are displayed differently, the buzzer sounds an alarm to show that the vibratory stimulation module is functioning properly.

In order to induce abnormal gaits in the participants, obstacles were placed on the path for the participants to negotiate. The obstacles were 10%, 20%, and 30% the participants' height. Five participants repeated the experiment twice for each obstacle, so a total of 30 passes were made, and in 28 of them, power was applied to the vibratory stimulation system. Power was not applied in two experiments with the 10% high obstacle. This is probably because the obstacle was low and thus the participants' gait was considered normal.

Through the two types of evaluation above, the accelerometer showed clearly the peak values of vertical acceleration for normal and abnormal gaits, and it was confirmed that the algorithm is useful in abnormal gait detection using peak values. In addition, we found that

the abnormal gait detection algorithm can detect abnormal gaits effectively.

Methods

In order to evaluate the effectiveness of the abnormal gait algorithm and the vibratory stimulation system, the participants
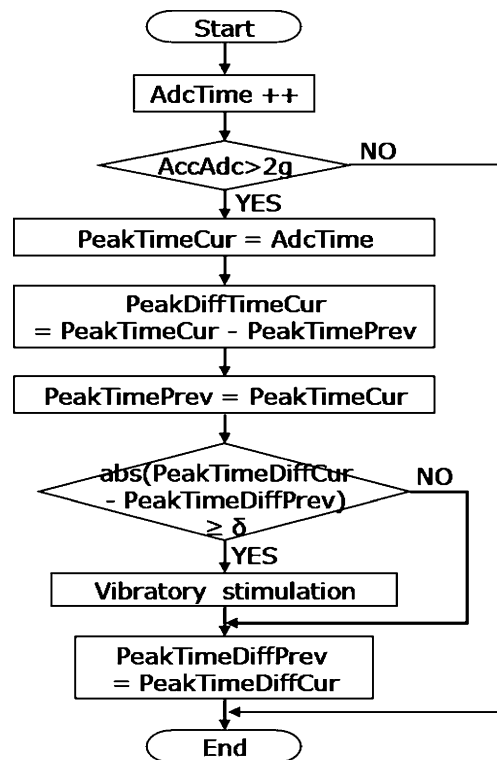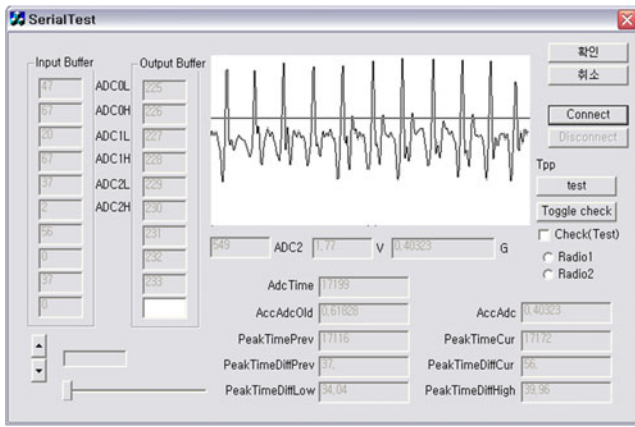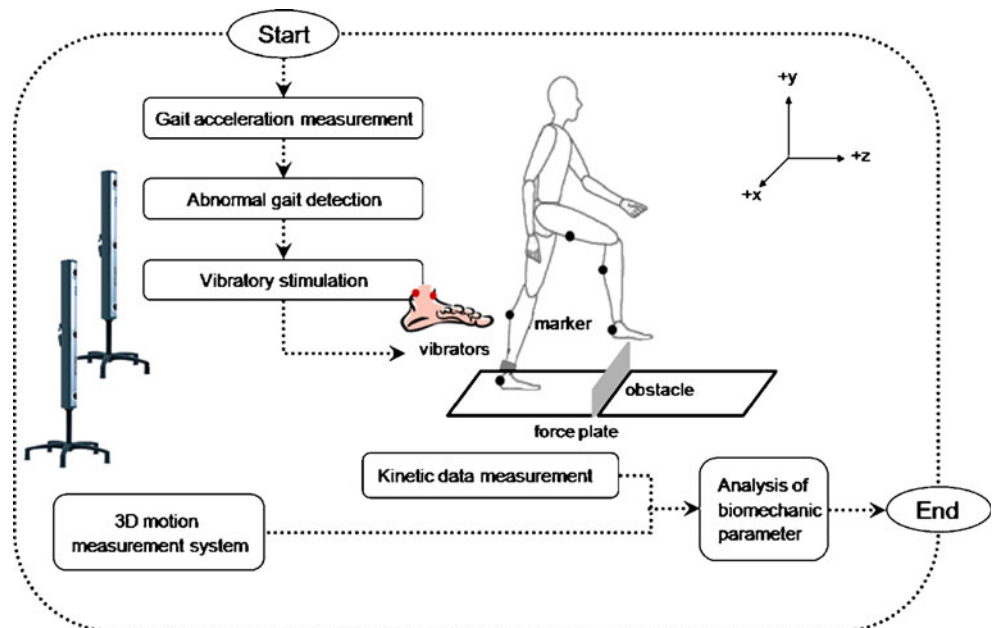

Fig. 4 Detection algorithm for abnormal gait

Fig. 5 Evaluation program of the algorithm

walked under varying vibration and obstacle conditions; kinematic changes were subsequently measured.

3D motion analysis system

In order to analyze the kinematic changes of participants during a gait, we used a three-dimensional motion analysis system (Optotrak Certus, Nothern Digital Inc., Canada). The system is shown in Fig. 6. The system is composed of two infrared cameras for observing markers attached to the body, two force plates to measure ground reaction force, several auxiliary devices for collecting data, and a computer for analyzing data. Based on anatomical reference points, we attached markers embedded with three infra-red light diodes (IREDs), one on the thoracic vertebra No. 12, one on the iliac crest, one on each of the thighs, one on each of the crura, and one on each of the ankle joints. The markers on the body were measured using two infrared cameras, and

sampled at a rate of 140 Hz. The three-dimensional axes were set as axis $+z$ for the walking direction, axis $+x$ for the right direction, and axis $+y$ for the vertical direction.

Vibratory stimulation system

The vibratory stimulation system used a vibration motor (JHV-10A1, JAHWA Electronics Co.) to apply vibratory stimulation to the participants' left ankle joint. The vibration motor is a flat-type, 3.4 ± 0.1 mm in height, 10 ± 0.1 mm in diameter, and 10 g in weight. The vibratory stimulation system is composed of vibration motors attached to a fixing band, each positioned on the left and right anterior tibial muscle and musculus triceps surae. Vibratory stimulation is generated for a second as power is applied to the vibration motor when the algorithm in the abnormal gait detection module detects an abnormal gait.

Experiment set-up

This study evaluated five adults in their 20s (four males and one female aged between 25~28). The participants did not have any neurological disease or disorder in their vestibular sense, and had normal musculoskeletal function. In addition, all of the participants had experience in experiments similar to this study, and before the experiment they were provided with sufficient information regarding the procedure, but not the purpose, of the experiment.

Participants had the gait measurement module attached to their right ankle joint, and the vibratory stimulation module a their left ankle joint. A gait was performed by stepping with the left foot on the first force plate, hurdling an obstacle, and then stepping with the right foot on the

Fig. 6 Schematic diagram showing the experimental paradigm

second force plate. To stabilize walking and the order of foot placement on the force plates, the participants had sufficient practice before the experiment. The experiment conditions were divided into gait without stimulation, and that with vibratory stimulation applied to the anterior tibial muscle and the musculus triceps surae. The obstacle was installed between the two force plates, and the obstacle condition (height) was 0%, 10%, 20%, or 30% of the participants' height. Each participant conducted the experiment twice for each of the two vibratory stimulation conditions x four obstacle heights, hence gait was measured a total of 14 times (except for the condition with vibratory stimulation at 0% height). In order to reduce the fatigue of the lower limbs resulting from the experiment, a ten minute break was given between the two experience types according to vibratory stimulation condition.

Data analysis

This study analyzed data collected using the three-dimensional motion analyzer and force plates, examined changes in the participants' kinematic parameters, and determined if vibratory stimulation was effective in stabilizing posture.

The measured parameters were: the difference between COM (center of mass) and COP (center of pressure) in the anterior–posterior (AP) direction, the displacement of COM in the AP direction, COM acceleration in the medio–lateral (ML) direction, and ground reaction force.

The difference between COM and COP in the AP direction is measured at the moment when the participant hurdles the obstacle. If the value is small, it means that the COM is close to the COP and the posture is stable. As shown in Fig. 7, the displacement of COP in the AP direction, the displacement of COM in the ML direction, and displacement of the right tiptoe in the AP direction are the differences between COM and position in each direction when the participant places the right foot on the ground before hurdling the obstacle, and COM and position in each direction when the subject steps the right foot on the ground after crossing the obstacle. COM acceleration in

the ML direction and the tilt of the trunk are measured at the moment when the participant crosses the obstacle. It is the difference in ground reaction force between when the participant places the left foot on the first force plate and when the right food on the second force plate.

Results

In order to see how human gait stability is affected by the abnormal gait detection and vibratory stimulation system developed in this study, we had the participants walk while wearing the vibratory stimulation system with different obstacle conditions. When the obstacle was 0 cm, the obstacle did not induce an abnormal gait, and therefore was excluded from comparison. Experiment data were statistically analyzed using SPSS 12.0, and statistical significance was tested with a p value <0.05.

Change in COP and COM in the AP direction

Figure 8 shows the difference between COM and COP in the AP direction. We compared results when the participant crossed four types of obstacles of different height, namely 0, 10, 20, 30% of the participants' height, when vibratory stimulation was applied and not applied. The difference between COM and COP in the AP direction increased with increased obstacle height, and tended to decrease when vibratory stimulation was applied. When comparison was made according to vibration condition, significant results were observed when the height of the obstacle was 10% ($p=0.001$) and 30% ($p=0.001$).

Figure 9 shows COM displacement in the AP direction with and without vibratory stimulation while hurdling the four obstacles of different heights. COM displacement in the AP direction increased with increased obstacle height, and showed a decreasing tendency when vibratory stimulation was applied. In comparison, according to vibration condition, the results were significant when the height of the obstacle was 10% ($p=0.006$) and 20% ($p=0.033$). A similar tendency was observed in the tilt of the

Fig. 7 Schematic diagram showing an overhead view of the walkway and the typical footfall pattern obtained from an obstacle trial
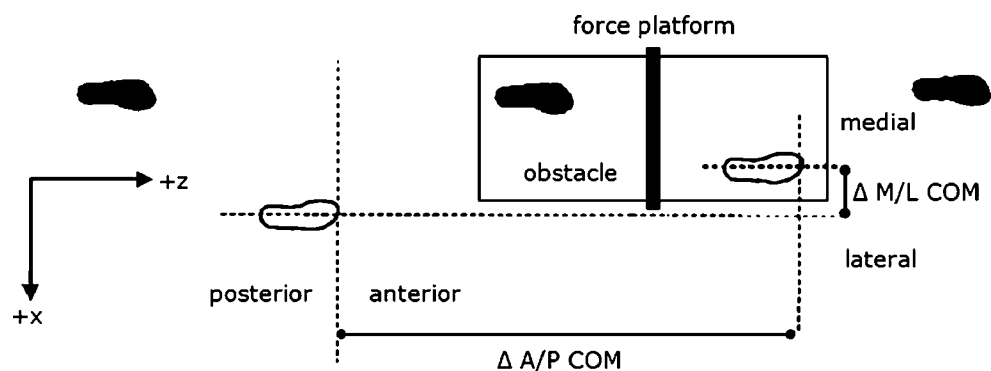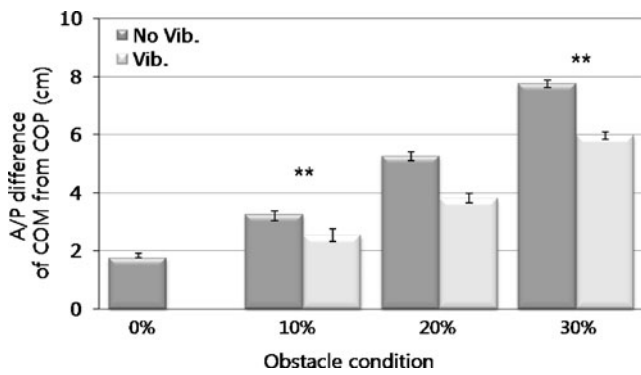
Fig. 8 Mean difference in the AP plane between COM and COP when the toe was over the obstacle for each vibration and obstacle condition (*p<0.05, **p<0.01)

Fig. 10 Mean trunk pitch when the toe was over the obstacle for each vibration and obstacle condition (*p<0.05)

trunk (Fig. 10). The tilt of the trunk increased with increased obstacle height, and showed a decreasing tendency when vibratory stimulation was applied. In comparison, according to vibration condition, the results were significant when the height of the obstacle was 20% ($p=0.001$) and 30% ($p=0.040$).

Figure 11 shows the displacement of the right foot in the AP direction, with and without vibratory stimulation while hurdling the four obstacles of different height. The displacement of the right foot in the AP direction increased with increased obstacle height, and showed a decreasing tendency when vibratory stimulation was applied. In comparison, according to vibration condition, the results were significant when the height of the obstacle was 10% ($p=0.024$), 20% ($p=0.018$), and 30% ($p=0.015$).

Change in COP and COM in the ML direction

Figure 12 shows the displacement of COM in the ML direction with and without vibratory stimulation system while hurdling the four obstacles of different height. The displacement of COM in the ML direction increased with
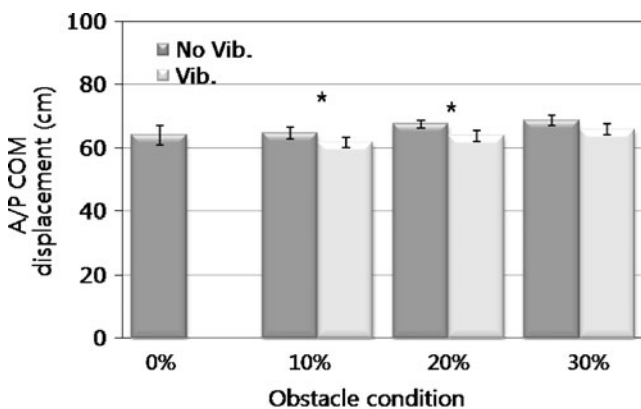
increasing obstacle height, and showed a decreasing tendency when vibratory stimulation was applied.

Figure 13 shows COM acceleration in the ML direction with and without power to the vibratory stimulation system while hurdling the four obstacles of different height. When the obstacle was high, COM acceleration in the ML direction increased, thus yielding an unstable posture. However, COM acceleration showed a decreasing tendency when vibratory stimulation was applied. In comparison, according to vibration condition, the results were significant when the height of the obstacle was 30% ($p=0.010$).

Figure 14 shows difference in ground reaction force when the participant places the left foot on the first force plate, and the right foot on the second force plate. We compared the results with and without power application to the vibratory stimulation system while hurdling the four obstacles of different height. The ground reaction force increased with increased obstacle height, and showed a decreasing tendency when vibratory stimulation was. The results were significant when the height of the obstacle was 30% ($p=0.049$).

Fig. 9 Mean A/P displacement of the COM during obstacle step over for each vibration and obstacle condition (*p<0.05)

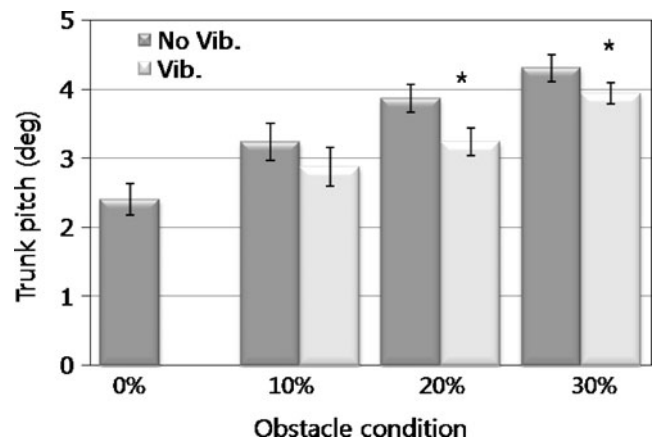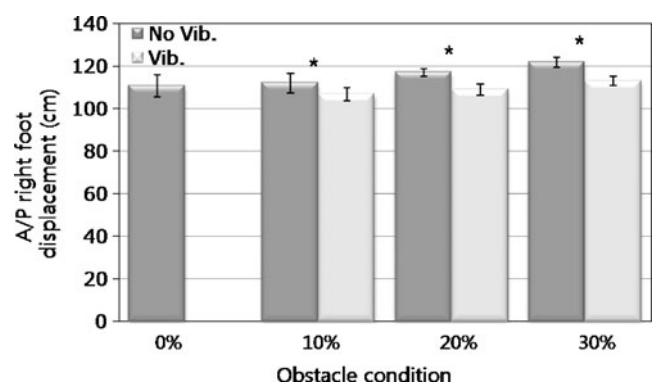Fig. 11 Mean A/P displacement of the right foot during obstacle step over for each vibration and obstacle condition (*p<0.05)

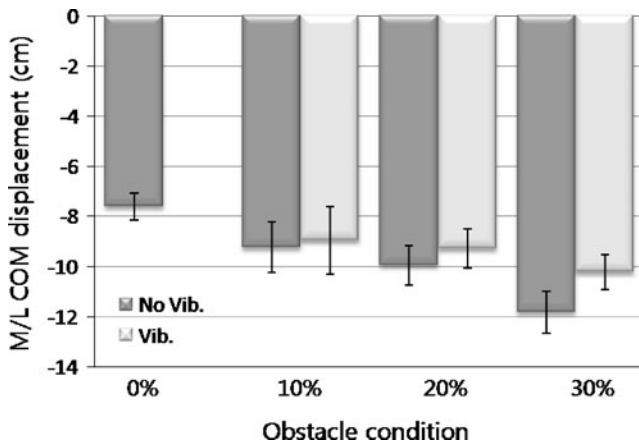Fig. 12 Mean medial–lateral displacement of the COM during obstacle step over for each vibration and obstacle condition

Fig. 14 Mean ground reaction upon stepping on the force platform (*$p<0.05$)

Discussion

The purposes of this study were to develop a vibratory stimulation system for improving posture stability and preventing falls, develop an algorithm to detect abnormal gaits and a vibratory stimulation system to prevent falls caused by abnormal gaits, and to evaluate their performance.

The present study developed an abnormal gait detection algorithm for improving gait stability using an acceleration sensor. Acceleration signals of human gait were first used in the fields of medical and motion analysis. Previous studies mainly used these signals for recognition of various human activities [9,10] such as walking, walking up/down stairs/slope, estimation of speed and incline of walking, evaluation of patient recovery, etc. Identifying users from gait signal for personal devices was first proposed by H. Ailisto [11]. Although in [11–12], other approaches of gait recognition which use acceleration data are described, our
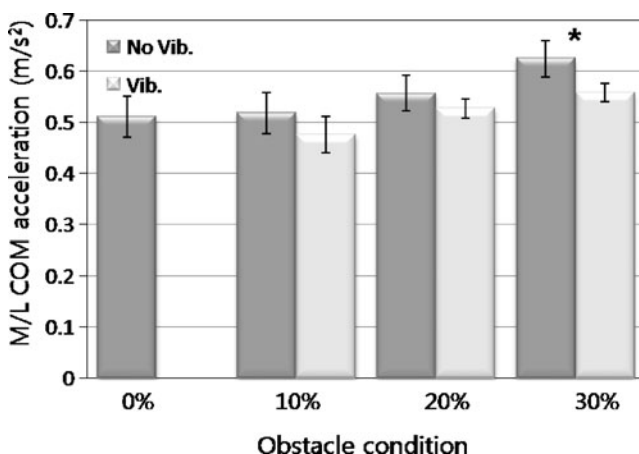
approach uses an accelerometer to measure the vertical acceleration of the ankle for evaluation of abnormal gait while negotiating an obstacle.

Vibration has become widely employed in studies of muscular proprioception. The sensitivity of muscle spindles to vibration stimulation depends on various mechanical characteristics, such as the intensity, displacement, frequency, and duration of the stimulus [13]. Vibration induces a stretch reflex, which is a muscle contraction in response to stretching within the muscle. The stretch reflex is a monosynaptic reflex, which provides automatic regulation of skeletal muscle length. When muscle lengthens, the spindle is stretched resulting in increased alpha motor neuron activity. Therefore, the muscle contracts and its length decreases as a result. The gamma co-activation is important in this reflex because it allows spindles in the muscles to remain taut, and therefore sensitive even during contraction [14]. Primary endings of ankle muscle spindles play a significant role in the control of posture and balance during gait by providing information on the movement of the body's center of mass with respect to the support foot [15]. Many studies have shown that vibration activates Ia afferents predominately [16]. Shinohara et al. [17] suggested that the increase in background EMG after vibration may indicate enhanced fusimotor drive to the muscle spindles (α–γ coactivation) to achieve the target force due to the vibration-induced decline in force capacity of the muscle. Alternatively, fusimotor drive to the muscle spindles may have been selectively enhanced after vibration [18–19]. We did not measure EMG, but measured ground reaction force as evidence of enhanced muscle activity after vibration. Ground reaction force increased with the obstacle height, and showed a decreasing tendency when vibratory stimulation was applied (Fig. 14). These results indicate that with increased obstacle height, the force supported by the left right feet increases gradually, and the application of vibratory stimulation equalizes these forces. Consequently, posture can be maintained after hurdling the obstacle.

Fig. 13 Mean medial–lateral acceleration of the COM during obstacle step over for each vibration and obstacle condition (*$p<0.05$)

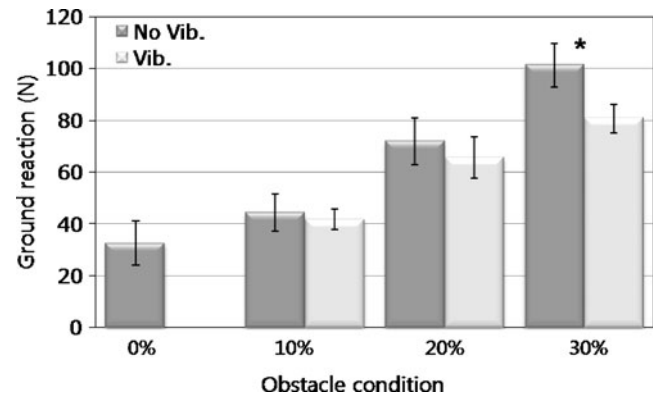We measured kinematic changes of participants during obstacle crossing to evaluate the abnormal gait and vibratory stimulation system. We then analyzed the changes in COP and COM in the AP and ML directions as parameters to evaluate the abnormal gait detection and vibratory stimulation systems.

The mean difference between COM and COP in the AP direction caused by vibratory stimulation applied to the ankle during obstacle negotiation were similar (Fig. 8). These results appear as a reflection controlled by the spine via plantar flexor activation. In addition, vibratory stimulation applied to the left ankle joint stabilizes the posture as COP inclines forward slightly more with regard to the left foot in the stance phase when the right leg hurdles the obstacle.

As the height of the obstacle increases, the body inclines forward and, as a result, the mean displacement of COM and trunk pitch in the AP direction both increase (Figs. 9 and 10). Vibratory stimulation applied to the left ankle joint is processed by the central nervous system and induces plantar flexor in the left foot. This means that according to changes in COP position by vibratory stimulation, compensational adjustment is made for restoring dynamic equilibrium, and through this mechanism, body weight is moved more stably. The displacement of the right foot position in the AP direction increased with increasing obstacle height, and showed a decreasing tendency with vibratory stimulation (Fig. 11). As the height of the obstacle increases, the body inclines forward and, as a result, the height and displacement of the right foot in the AP direction also increase. Vibratory stimulation applied to the left ankle joint is processed by the central nervous system and induces plantar flexor in the left foot. Through this mechanism, the position of COM is moved backward, and the displacement of the right foot decreases.

For the changes in COP and COM in the ML direction, we measured the displacement and acceleration of COM as parameter to evaluate the abnormal gait detection and vibratory stimulation systems. When the obstacle is high, COM inclines outside slightly more, and the inclination is reduced by the application of vibratory stimulation (Fig. 12). If vibration is applied to the left ankle joint, the foot is placed inside slightly more with regard to COM, and thus the step becomes shorter. This is the result of compensation for the wrong interpretation by the central nervous system that COM moves toward slightly more inside due to change in the muscle spindles induced by the vibration. That is, change in COM in the ML direction is induced by changing the activity of the muscle spindles. As a result of the vibration on the left ankle joint, the right foot is placed slightly more inside with regard to COM thereby yielding a shorter step. As a result, right foot becomes closer to the left leg and COM acceleration in the ML direction decreases (Fig. 13).

Previously it has been demonstrated that during normal straight path walking there is a significant relationship between COM-COP (COM position relative to the COP) and COM acceleration in both the frontal and sagittal planes [20]. The body acts as an inverted pendulum and the distance of the COM from the base of support determines the rate at which the body falls under the influence of gravity [21]. A few studies have reported the adequacy of the COM-COP interaction in demonstrating dynamic stability (balance control during locomotion), with a consistent COM trajectory passing between the alternating COP of each supporting foot [21–22]. Recent results showed that COM ML displacement and peak ML velocity during obstacle crossing could be used to better detect dynamic instability in elderly adults [13–24]. In the present study, we observed vibration-induced changes to COM-COP that were not accompanied by corresponding changes in COM acceleration and vice versa. However, it should be noted that during walking, foot placement is the primary determinant of COP position. Although COP position within the stance foot is likely to change in response to contraction of ankle muscles, the resulting change in COM-COP is unlikely to be large enough to make a significant change to COM acceleration. Our results clearly show that there were significant changes in COM-COP in response to posterior and anterior ankle muscle vibration attributable to a significant posterior shift in COP location due to ankle plantar flexion. Although this vibration-induced change in ankle rotation was large enough to cause detectable changes in COP location, it was not large enough to produce significant changes to COM behavior.

The fact that we did observe vibration-induced changes in M/L COM acceleration without changes in COM-COP suggests that the stance ankle muscle inversion/eversion does not result in significant changes to M/L COP location. While the relationship between COP-COM and COM acceleration is significant during walking, it is not perfect due to the nature of the body segment movements. This lack of a perfect inverted pendulum during locomotion may result in observed changes in COM acceleration, but may not be reflected in the COP-COM measure.

The risk of falling is determined by the frequency of imbalance episodes and the ability to recover balance [25]. While it appears there is no convincing evidence that older adults experience a greater probability being tripped [26], the capacity to recover balance from an imbalance episode is markedly reduced in older persons. Results to this effect have been reported for a wide range of fall-inducing perturbations [27–29]. The focus of previous studies regarding the risk of falling is the decline of muscle strength. Isometric strength of the ankle muscles declines with age at a rate of about 1–1.5%/year after the

fifth decade [30]. The main cause of age-related strength reduction is reduced muscle mass (termed senile sarcopenia) due to loss of muscle fibers (apoptosis) and a decrease in muscle fiber size (atrophy) [31–32]. Therefore, we developed a system of vibratory stimulation on a lower limb for use when an abnormal gait is generated due to imbalance episodes. An accelerometer shows the peak values of vertical acceleration during normal and abnormal gaits, and can be useful in abnormal gait detection. Moreover, the abnormal gait detection algorithm could detect abnormal gaits effectively, and provide immediate feedback on abnormal gaits using vibratory stimulation (detection accuracy: 95%). Vibratory stimulation applied to the anterior tibial muscle and musculus triceps surae in a dynamic state provides immediate feedback on abnormal gaits and stabilizes posture effectively.

In order to develop a vibratory stimulation system for improving posture stability and preventing falls, this study developed an algorithm to detect abnormal gaits and a vibratory stimulation system to prevent falls caused by abnormal gaits, and then evaluated their performance. From the results of this study as presented above, it was concluded that the abnormal gait detection and vibratory stimulation system proposed in this study is effective in stabilizing posture while walking. If the system is used while walking by older or weak patients whose muscle strength in the lower limbs is weak, it is expected to stabilize their posture and prevent falls.

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