# Comparative Analysis of Walking Gait Cycle between Healthy People and Walking Disabilities to Prevent Tripping Using Wearable Device and KINECT

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*Abstract*— **Elderly people, especially walking disabilities, have an increased risk of falling and consequently injuring themselves. They need to be prevented from falling to maintain their health because injuries from falling are a major reason for them to be hospitalized or placed in residential care. Motion capture systems are a key component to prevent falls. We comparably analyzed walking gait cycle between healthy people and walking disabilities using a wearable device (WD) and KINECT to detect warning signs of falls. In this paper, we experimentally clarify what signs are useful to prevent falls. We developed a gait monitoring device comprised of a smartphone application and a pair of shoes on which WDs are mounted to measure such warning signs, and proposed presentation formats for data measured by KINETC.**

***Keywords-falls; trip; hemiplegia; gait; shoes; MS-KINECT.***

## I. INTRODUCTION

The authors comparably analyzed walking gait cycles between healthy people and walking disabilities using gait monitoring shoes on which wearable devices (WDs) were mounted [1].

As the percentage of elderly people in the populations is increasing around the world [2], the number of functionally impaired people, such as cerebrovascular patients who are paralyzed down one side, will also increase. These people have an increased risk of falling and consequently injuring themselves [3][4]. Falling down is one of the main reasons for them to be hospitalized or placed in residential care.

There are many studies on falls by elderly people. The World Health Organization Regional Office for Europe analyzed these studies and classified fall risks amongst elderly people by history of falls, age, gender, living alone, ethnicity, medicine, medical conditions, impaired mobility and gait, sedentary behavior, psychological status - fear of falling, nutritional deficiencies, visual impairments, and foot problems [4]. Stroke patients, such as those with cerebrovascular disease, especially are at a substantially high risk of falling [5][6][7][8]. Their higher frequency of falls is due to weak muscles, one-side paralysis, and downward-pointing toes. For people with impaired mobility and gait, tripping is a major cause of falls [9][10], so we focus on tripping in this paper.

Since weak muscles, one-side paralysis, and downward-pointing toes strongly appear in the movement of legs and feet, motion capture for them is a key component to analyze impaired mobility and gait, and useful to prevent tripping, and conducts therapy and rehabilitation of hemiplegia.

Here, we focus on extracting warning signs of tripping for walking disabilities, such as cerebrovascular patients. In this paper, a WD is mounted on a shoe to measure the acceleration and angle velocity, and Microsoft KINECT [11] is used to measure positions of each joint of the lower body.

We obtained output data of an acceleration sensor and gyroscope sensor in a WD, Sony Smart Watch 3, mounted on the front part of a shoe to estimate the kicking power and change of angle between a foot and the floor. We noticed that the angle velocity at the terminal stance and the angle at the terminal swing are clearly different for unimpaired subjects and walking disabilities such as stroke. Moreover, they clearly have different step lengths as measured by KINECT. We also developed a monitoring device comprised of a smartphone application and a pair of shoes on which WDs were mounted, and proposed using the side and top view formats to present data measured by KINECT.

After introducing related works in Section II, we consider how people trip on a flat floor in Section III. Different features between physically unimpaired students and walking disabilities such as stroke are extracted from measured data in Section IV. Gait monitoring shoes and monitoring application, and gait presentation format with a KINECT are introduced in Sections V and VI. Measuring and analyzing a walking gait for walking disabilities are described in Section VII. Finally, conclusions are summarized in Section VIII.

## II. RELATED WORKS

In this section, we introduce motion capturing devices.

### A. *Sensor usage type*

Weijun Tao et al. reviewed gait analysis technologies based on wearable sensors that were the accelerometer, gyroscope, electromagnetic tracking system, magneto-resistive sensors, flexible goniometer, sensing fabric, force sensor, and so on [12]. They mentioned that fall risk estimation is an important application of gait analysis using wearable sensors. However, they did not describe about motion of gait for elderly people or walking disabilities.

Stacy J. Morris Bamberg et al. developed a prototype shoe in which several kinds of wearable sensors, such as accelerometer, gyroscope, force sensor, bidirectional bend sensor and so on [13]. The calibrated sensor outputs were almost same as results obtained simultaneously from a biological motion measuring equipment. They calculated the maximum pitch (angle between the shoe sole and floor at the toe-off timing), minimum pitch (angle between the shoe sole and floor at the heel-strike timing), the stride length from output of accelerometers and gyroscopes. They also compared the maximum pitch, minimum pitch and stride length between the healthy gait and parkinsonian gait. There were differences on mean value of calculated data between the healthy gait and parkinsonian gait. However, considering standard deviation of calculated data, such differences were small. They also did not measure and analyze motions of gait for elderly people or walking disabilities.

Farzin Dadashi et al. measured motion of gait for many elderly people with shoe-worn inertial sensors and provided normative values for a clinician to measure reference gait parameters [14]. They analyzed motion of gait and clarified the difference in gait parameters, such as the clearance between a shoe sole and floor, gait speed, stride length between males and females by considering the effect of age factors. However, their data did not show differences clearly between the male and female, and the effect of age factor. And, they did not investigate data for walking disabilities or analyze reasons for tripping.

Mourad Benoussaad et al. introduced a method to robustly estimate foot clearance during walking using a single inertial measurement unit (IMU) placed on the subject's foot [15]. In their paper, the foot clearance was the height of ankle from a floor. However, the toe clearance is more critical for tripping. And, they did not measure the toe clearance for walking disabilities such as stroke and analyze reasons for tripping.

## B. *Camera usage type*

Vicon is one of the most famous companies in the motion capture industry. They can measure complex motions of joints in a body [16]. Vicon's system needs plural specialized video cameras, and know-how is needed to measure motions of joints. Thus, this system is too expensive for a small rehabilitation center or an individual to purchase and operate.

KINECT is one of motion capture devices distributed by Microsoft [11]. Since its price is a few hundred dollars, it is possible for small rehabilitation facilities to introduce it. There are many researches that use KINECT. Obdrzalek et al. compared the Kinect pose estimation with more established techniques relying on motion capture data [17]. They said that system such as Kinect has significant potential as a low-cost alternative for real-time motion capture and body tracking in health applications. We also used KINECT for a remote rehabilitation system of which content was a standing-up training [18]. In this paper, three kinds of view method, that are the front-view, side-view and top-view, were introduced to present a strain of the upper body.

# III. CONSIDERATION OF TRIPPING FACTOR

When the swing foot progression is unexpectedly obstructed, a trip occurs that leads to a forward rotation of the body and eventually might cause a fall.

Mourad Benoussaad et al. measured the minimum toe clearance (MinTC) to avoid tripping [15]. MinTC is a critical value to clear obstacles on the ground or floor. However, elderly people, especially those who have had strokes, sometimes trip on flat ground or floors, not obstacles. In this section, we consider reasons a person trips on flat ground or floors. We divide the normal walking gait cycle into eight phases the same as Weijun Tao et al. as shown in Fig. 1 [12]: (1) initial contact (heel-strike timing), (2) loading response, (3) mid-stance, (4) terminal stance (toe-off timing), (5) pre-swing, (6) initial swing, (7) mid-swing, and (8) terminal swing.

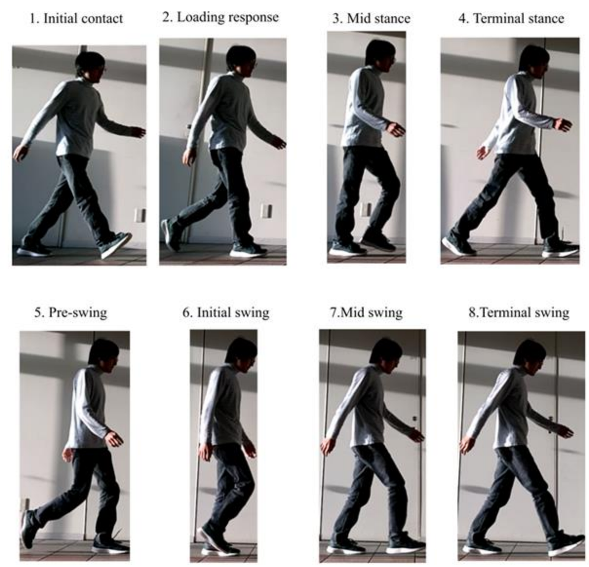



Figure 1. Normal walking gait cycle (See a right foot)

Most walking disabilities have weak muscles and are hard to raise their toe. They are at risk of three types of trips.

- Case 1: A toe touches the floor first instead of a heel at phase 1. Since phases 2-5 are skipped, the toe is dragged along the floor. When the dragging strength is stronger than the person's muscular power, he/she trips (Fig. 2(a)).
- Case 2: Kicking power of the front part of a foot is insufficient at phases 4 and 5 to raise the heel and toe up from the floor. In this case, a person does not swing but shuffles. When the frictional force between a shoe sole and the ground or floor is stronger than his/her muscular power, he/she trips (Fig. 2(b)).
- Case 3: A toe touches the floor due to it pointing down during the swing phases (5-8), and the knee goes further forward than the foot. When the dragging strength is stronger than the person's muscular power, he/she trips (Fig. 2(c)).

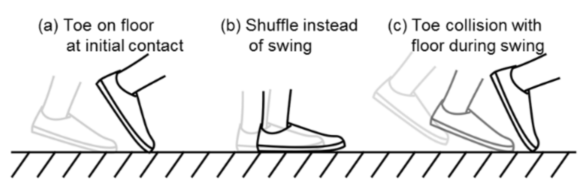



Figure 2. Cases of tripping

The above problems suggest that the kicking power at phases 4 and 5 and the angle between the foot and a floor are critical parameters.

## IV. EXTRACTION OF WARNING SIGNS FOR TRIPPING

In this section, we experimentally investigate whether the kicking power at phases 4 and 5, and the angle between the foot and lower limb are critical parameters.

### A. Experimental method

Since kicking power must be expressed as the angle velocity or the acceleration for the foot, we mounted a WD which had an accelerometer and gyroscope on the foot. In this experiment, we used Sony SmartWatch 3 as a WD which is mounted on the front part of a foot with Velcro tape as shown in Fig. 3. This mounting position was same as one in Farzin Dadashi's experiment [14]. The sampling rate was 40 msec.

We measure angle velocity for up and down directions of the front part of the foot (X axis of a 3D gyroscope). We also adopted a three-point moving average of the angle velocity to calculate the angle, because output values extremely change up and down. Therefore, the angle for X axis $Angle_{xn}$ at time $t_n$ is calculated as follows.

$$Angle_{xn} = Angle_{xn-1} + \frac{t_n - t_{n-1}}{1000} \times \frac{G_{xn-1} + G_{xn} + G_{xn+1}}{3} \quad (1)$$

$G_{xn}$ is the value of angle velocity for X axis at time $t_n$.

We investigated the measuring accuracy of Sony SmartWatch 3 using a slant rule as shown in Fig. 4. We measured data five times. Calculated angles vs. angles given by the slant rule are listed in Table I. These data showed calculated angles were so accurate. We noticed drift errors of a gyroscope that increase the value by 0.2 rad. /sec. during a WD sets on a flat floor. However, each measurement lasted less than 20 sec. Therefore, we think the effect of the drift error is negligible.

(a) WD: Sony SmartWatch 3 (b) WD mounted on foot

Figure 3. Measuring device and WD mounting method


Figure 4. Slant rule

TABLE I. ACCURACY OF CALCULATED ANGLES

| Given angle (degree) | Calculated angle (degree) | Standard deviation (degree) |
|---|---|---|
| +50 | +49.00 | 0.45 |
| +40 | +39.17 | 0.74 |
| +30 | +28.92 | 0.51 |
| +20 | +19.58 | 0.63 |
| +10 | +9.21 | 0.58 |
| 0 | 0.21 | 0.15 |
| -10 | -9.72 | 0.58 |
| -20 | -19.47 | 0.34 |
| -30 | -30.70 | 0.56 |
| -40 | -40.57 | 0.41 |
| -50 | -50.54 | 0.53 |

We also measured the foot stride (FS) and recorded motions of subjects' knees, ankles, and feet by using KINECT. The UNIX time was introduced to synchronously measure data with a WD and KINECT.

We defined the FS as the maximum difference between positions of the right and left foot measured by KINECT as shown in Fig. 5. We experimentally looked for the height of KINECT to measure accurately. As the result, the height of KINECT is 75 cm. Moreover, we set the face angle of KINECT so that "+" markers on the display are superimposed on "-" markers on the floor to correct KINECT as shown in Fig. 6. From the bottom of the image in Fig. 6, these markers correspond to 2, 3, 4, and 5 m from KINECT. Since this picture is output of KINECT's video camera, left and right are reversed.

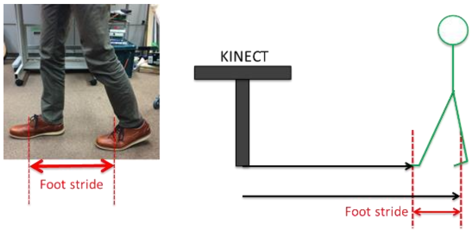



Figure 5. Definition of the foot stride (FS)

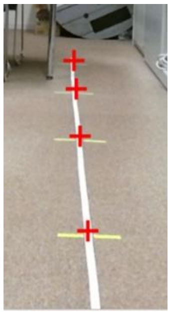
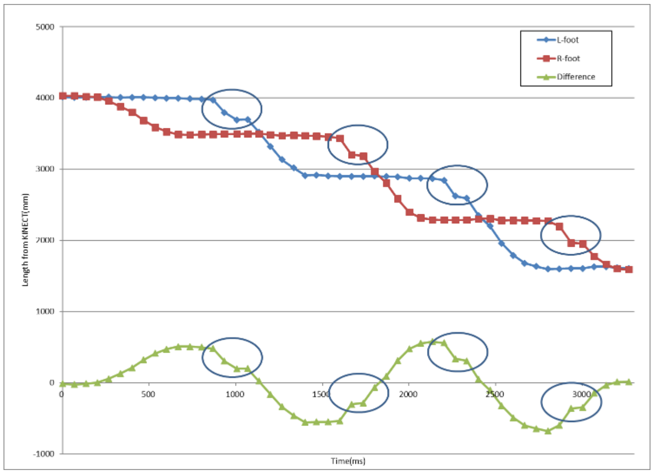
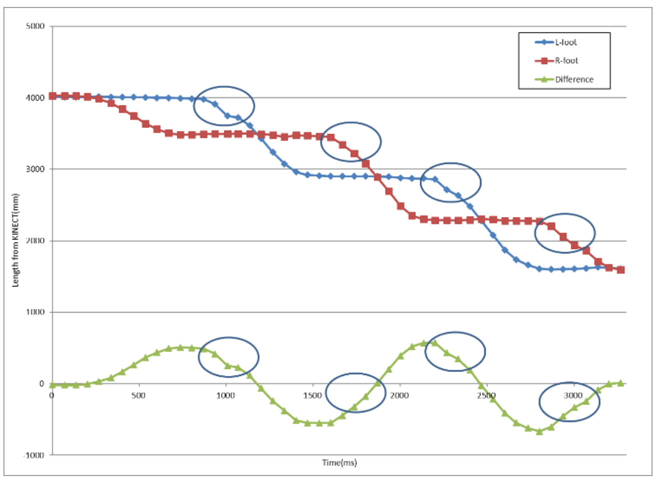


Figure 6. Pre-setting of KINECT

Fig. 7 (a) shows distance from KINECT for a left and right foot measured by KINECT. The blue line shows the distance between the left foot and KINECT, the red line shows the distance between the right foot and KINECT, and the green line shows the length for the left foot minus the length for the right foot. The minimum and maximum values of the green line very clearly correspond to FS for each step. Three-point moving average curves for measured data are shown in Fig. 7 (b) for reference. Circled parts correspond to movement of the feet from the mid-stance to terminal stance. Since the three-point moving average masks such feet motions, we decide to present a graph containing raw data.




(a) Raw measured data




(b) After processing with three points moving average

Figure 7. Measured length between KINECT and the left or right foot

We experimentally evaluated the accuracy of measured FSs. Participants were eight unimpaired university students. They walked on three sets of stride markers of 30, 60, and 70 cm as shown in Fig. 8.

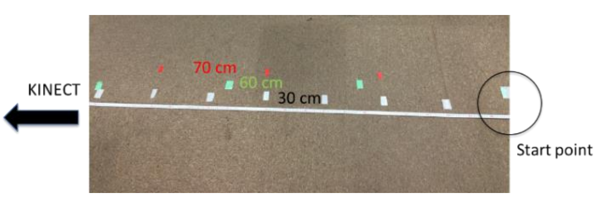



Figure 8. Three pitches markers for the accuracy of measured FS

Experimental results are shown in Table II. Since the range within which KINECT can definitely measure is between 1.5 and 4.5 m, average values in Table II are averages of absolute values for strides between them. Errors were less than 10%, and the standard deviations were less than 3 cm. We evaluate the errors within allowance.

TABLE II. ACCURACY OF MEASURED FS

| Pitch (cm) | Average (cm) | Standard deviation (cm) |
|---|---|---|
| 30 | 27.7 | 2.02 |
| 60 | 54.2 | 2.65 |
| 70 | 63.1 | 2.67 |

### B. *Measured data and consideration*

We measured the acceleration, angle velocity and angle for five physically unimpaired students and three walking disabilities using a WD as same as the former sub-section. Every walking disabilities in this experiment had one-side paralysis, and trained periodically at a rehabilitation facility. Some of them used a wheel chair and could not walk by himself before training. They walked along a straight line to MS-KINECT. A WD was attached on the front part of foot on the paralysis side as shown in Fig. 9. We measured data for each patient two times.

(a) MS-KINECT set up in rehabilitation facility

(b) WD mounted on a foot of patient

Figure 9. Measurement environment in rehabilitation facility

Figs. 10 and 11 show examples of change of acceleration, angle velocity, and angle for a physically unimpaired student and a walking disability. Data for two steps are plotted.

Each flat period (roughly the center period) in these figures is when the entire shoe sole touched the floor; this period corresponds to phases 2 (loading response) and 3 (mid-stance). The reason that the value during this period is not zero is that the WD measures the angle between the front part of the foot and the floor, which depends on the person and shoe. Therefore, we reset this angle for the gait monitoring shoes described in Section V when the entire shoe sole touched the floor. This processing enables the WD to measure the angle between the back of the foot and the floor, and removes the drift error of the gyroscope. This value does not depend on person or shoe.

The maximum angle velocity at timing A means the kicking power from phase 4 (terminal stance) to 5 (pre-swing), and the minimum angle at timing B means the angle to the floor at phase 8 (terminal swing).

Lower angle velocity at A in Fig. 10 is about 420 deg./sec. On the other hand, higher angle velocity at A in Fig. 11 is about 250 deg./sec. Thus, a physically unimpaired student and a walking disability obviously differ in terms of gait. The walking disability clearly has weaker kicking power at phase 4 (terminal stance) than the physically unimpaired student.

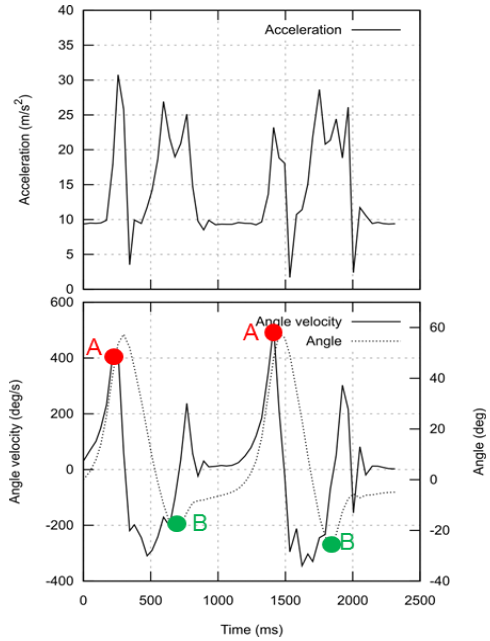



Figure 10. Changes of angle velocity, angle, and acceleration for physically unimpaired student

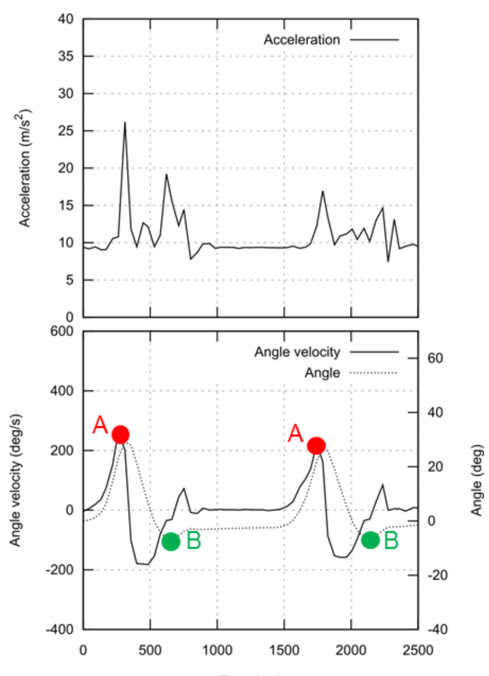



Figure 11. Changes of angle velocity, angle, and acceleration for walking disability

Higher angle at B in Fig. 10 is about -18degree. On the other hand, lower angle at B in Fig. 11 is about -8degree. Thus, a physically unimpaired student and a walking disability obviously differ in terms of the angle to a floor at phase 8 (terminal swing). This shows that it is difficult for a walking disability to raise his or her toe at the terminal swing phase.

The other hand, the acceleration basically changes corresponding to the angle velocity and angle. However, they have much noise, and their amplitudes are not stable.

Tables III and IV list the averages and standard deviations (SDs) of measured data for angle velocity at timing A and angle at timing B. The angle velocity at timing A is clearly different between unimpaired students and walking disabilities. There is a big difference between them in the angle at timing B, however, this value would have sometimes overlapped each other.

Table III. Angle velocity at the terminal stance

| Participant | Average (deg./s) | SD (deg./s) |
|---|---|---|
| Student | 509.36 | 18.91 |
| Walking disability | 342.06 | 86.52 |

Table IV. Angle at the terminal swing

| Participant | Average (deg.) | SD (deg.) |
|---|---|---|
| Student | -17.76 | 8.02 |
| Walking disability | -7.45 | 8.02 |

We also measured the FS using KINECT, and the cadence for a gait using a WD and KINECT. Table V lists the averages and SDs of measured data for the strides. In this paper, we define the cadence as the number of steps per minute. We estimated the cadence derived from an average of 10 intervals between one timing A and the next A, which were peak angle velocities of a step, when a WD was used. Estimated cadences are listed in Table VI.

There are clearly differences between unimpaired participants and walking disabilities in terms of the FS. FSs of walking disabilities are more than 10 cm shorter than those of unimpaired participants. On the other hand, the cadences of walking disabilities are slightly faster than those of unimpaired participants. Most physiotherapists said that FSs of elderly people, especially walking disabilities, are usually shorter than those of unimpaired people. These data prove what physiotherapists know experimentally.

TABLE V. FOOT STRIDE (FS)

| Participant | Average (cm) | SD (cm) |
|---|---|---|
| Student | 60.0 | 6.5 |
| Walking disability | 42.6 | 1.8 |

TABLE VI. ESTIMATED CADENCE

| Participant | Average (steps/m) | SD (steps/m) |
|---|---|---|
| Student | 46.4 | 5.0 |
| Walking disability | 49.0 | 6.2 |

On the basis of the results of these experiments, **we decided to adopt the angle velocity at the terminal stance to initial swing, angle between a foot and floor at the terminal swing, and average FS to detect warning signs of falls.** Section V introduces a pair of shoes and smartphone application to measure angle and angle velocity, and Section VI shows presentation formats for data measured by KINECT.

## V. GAIT MONITORING DEVICE

### *A. Shoes*

A WD has to be attached somewhere on a body during walking to detect signs of tripping to prevent a fall. A WD was attached to the front part of the foot in Section IV. However, it is difficult for a WD to firmly be set at this place for a long time because it is easily detached. Therefore, we studied which position is the best to detect the change of angle velocity for a foot and angle between a foot and floor. We attached WDs to a heel and a lower limb as shown in Fig. 12.

For this test, we used STEVAL-WESU1 by STMicroelectronics (see Fig. 13) as a WD instead of Sony SmartWatch 3. This wearable unit includes four sensors:

- 3D-accelerometer,
- 3D-gyroscope,
- 3D-magnetometer,
- MEMS pressure.

This device is 37 x 40 x 8 mm and weighs 9.6 g.

We inserted STEVAL-WESU1 into the heel of a shoe as shown in Fig. 12 (a) (details in the next sub-section). Angle velocity and acceleration data of STEVAL-WESU1 are sent to and processed by an Android smartphone. The sampling rate was 40ms. We adopt a three-point moving average to remove noise.

(a) Heel

(b) Single limb

Figure 12. WD attaching position

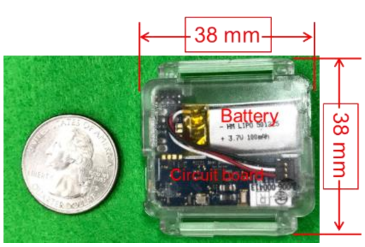



Figure 13. STEVAL-WESU1 by STMicroelectronics

We requested three unimpaired students to walk with their normal gait. Since their data change was basically the same, graphs of one participant are shown in Figs. 14 and 15. Both plotted lines in Fig. 14 are similar in shape to those in Fig. 10. Timing A and B correspond to timing A and B in Fig. 10. Timing B in Fig. 14, which is the angle at the terminal swing, is shown more clearly than that in Fig. 10. On the other hand, timing C in Fig. 15 shows the kicking power from phase 4 (terminal stance) to 5 (pre-swing) is the same as timing A in Figs. 10 and 14. However, the angle at timing D in Fig. 15 is between not the foot and floor but a single limb and the vertical line to the floor. The plotted angle in Fig. 15 clearly shows a change of angle for the single limb.

As the result of this experiment, we decided that the heel was the best position to place a WD.

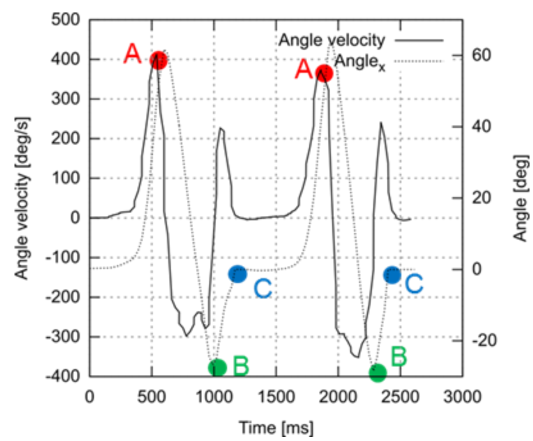



Figure 14. Angle velocity and angle data at heel in normal walk

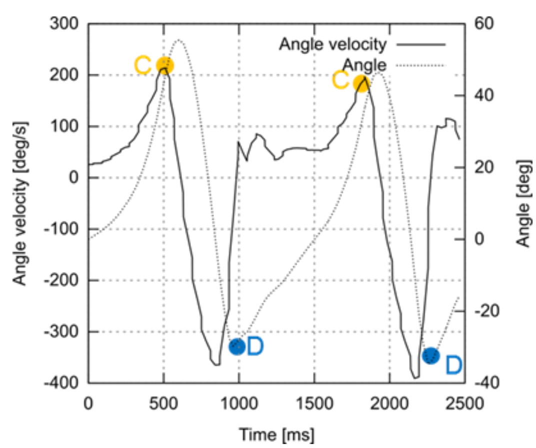



Figure 15. Angle velocity and angle data at single limb in normal walk

### *B. Monitoring application for smartphone*

As described in former sub-section, we determined the heel of a shoe is the best place to measure angle velocity of the foot and angle between the back part of a foot and the floor. We inserted a WD (STEVAL-WESU1 by STMicroelectronics) into soles of both shoes. And, we also developed a gait monitoring application for Android smartphone which measures and stores the angle velocity and angle as shown in Fig. 12. The upper part shows ID of WD for the right and left shoe, and the lower part shows angle velocities at A in each step for right foot, angle at B in each step for right foot, angle velocities at A in each step for left foot, and angle at B in each step for left foot. In this application, direction of angle is turned. When these graphs were measured, a participant played a stroke patient who had a one-side paralysis for the right side of the body. Therefore, most strength of angle velocity at A for right foot were smaller than that for left foot. And, most amplitude of angle at B for right foot were smaller than that for left foot.

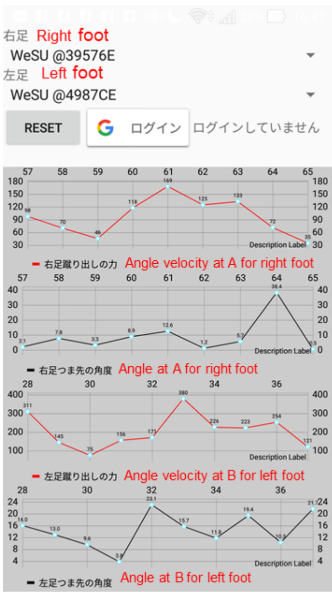
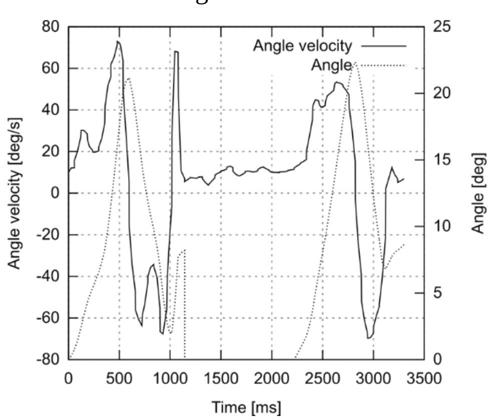


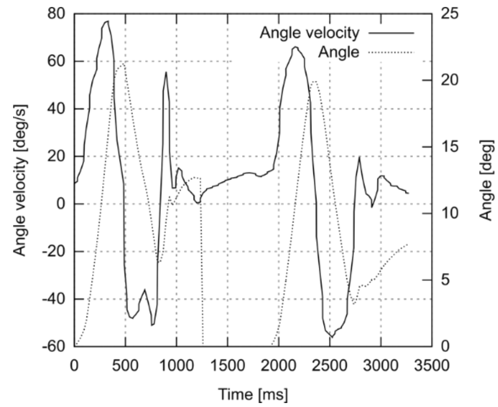

Figure 12. Gait monitoring application for Android

## C. Measured data using gait monitoring shoes

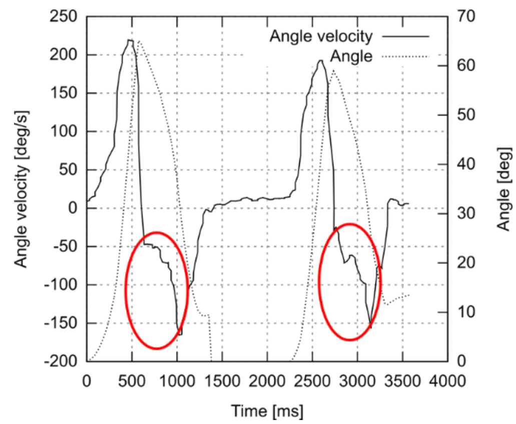
We experimentally monitored the walking gait for two participants. They were unimpaired people. They walked and played the three types of trips in Fig. 2. Example measured data are shown in Fig. 13. Therefore, curves of angles in these graphs have different discontinuity to those in other graphs at the sole of a shoe touching a floor.




(1) Toe on floor at the initial contact




(2) Shuffle instead of swing



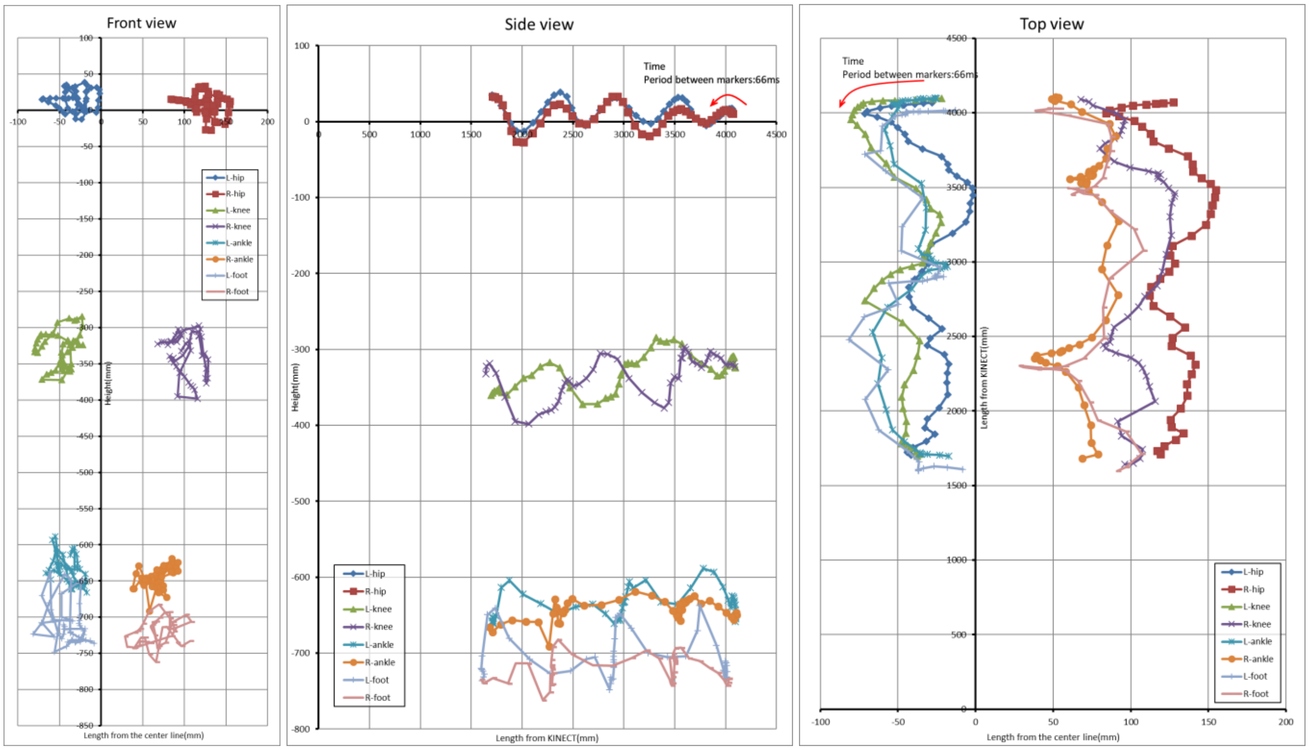

(3) Toe collision with floor at the initial swing

Figure 13. Example measured data for tripping with gait monitoring shoes

In (1); toe touching the floor first instead of a heel, and (2); shuffling, shapes of angle velocity resemble that of the normal walk shown in Fig. 10. However, maximum values of angle velocity and minimum angle in a cycle in Fig. 13 (1) and (2) are much smaller than those in Fig. 10. Their absolute minimum values are also much smaller than those in Fig. 10. This feature must show that when muscle strength is weaker, more trips occur. The red circle in Fig. 13 (3) shows this situation clearly. In the case of a normal walk, angle velocity rapidly decreases from the pre-swing to the initial swing. However, in (3), the angle velocity limply decreases on the way.

## VI. GAIT PRESENTATION FORMAT FOR KINECT

The above gait monitoring device is useful to measure degrees of muscle power and the angle between the foot and floor. However, they have difficulty measuring the position of joints of the lower body such as the foot and ankle. Hence, we measured them using KINECT. Since we noticed that three kinds of view image (front, side, and top views) were useful to find out the strain condition of the upper body [18], we adopted these three view graphs for the gait cycle. Before measuring joints of the lower body for walking disabilities, we measured them for an unimpaired person to evaluate




(a) Front view (b) Side view (c) Top view

Figure 14. Positions of joints on the lower body for natural gait

KINECT's measurement accuracy. The walking course was the line shown in Fig. 6. A participant walked in three ways:

(1) Natural gait: a participant puts one foot on either side of the center-line (Fig.14).
(2) On-line gait: a participant puts his/her foot on the center line (Fig.15).
(3) Circumduction gait: a participant moves his right foot naturally and exaggeratedly rotates his/her left-foot away from the center-line (Fig.16). Some hemiplegia patients move their palsied foot with this walking form.

In Figs. 14 to 16, the original position is the center of KINECT for each direction. Since KINECT is set 75 cm above the floor, height of the floor is -75 cm. Data for hips, ankles, knees, and the front part of feet are presented in these figures.

The front-view shows moving height ranges for each joint. However, it is impossible to detect the position of each joint of the basis of walking steps. On the other hand, the side and top views respectively show the change of each joint in the vertical and horizontal directions in accordance with walking. The change of the moving height range of hips is very clearly shown in all side view graphs. However, measured data for the heights of knees, ankles, and feet by KINECT did not change smoothly. Since lengths from KINECT for both feet were very accurately measured as shown in Fig. 7, we decided not to adopt the moving average processing. Hence, we will consider whether some processing should be adopted to smooth them.

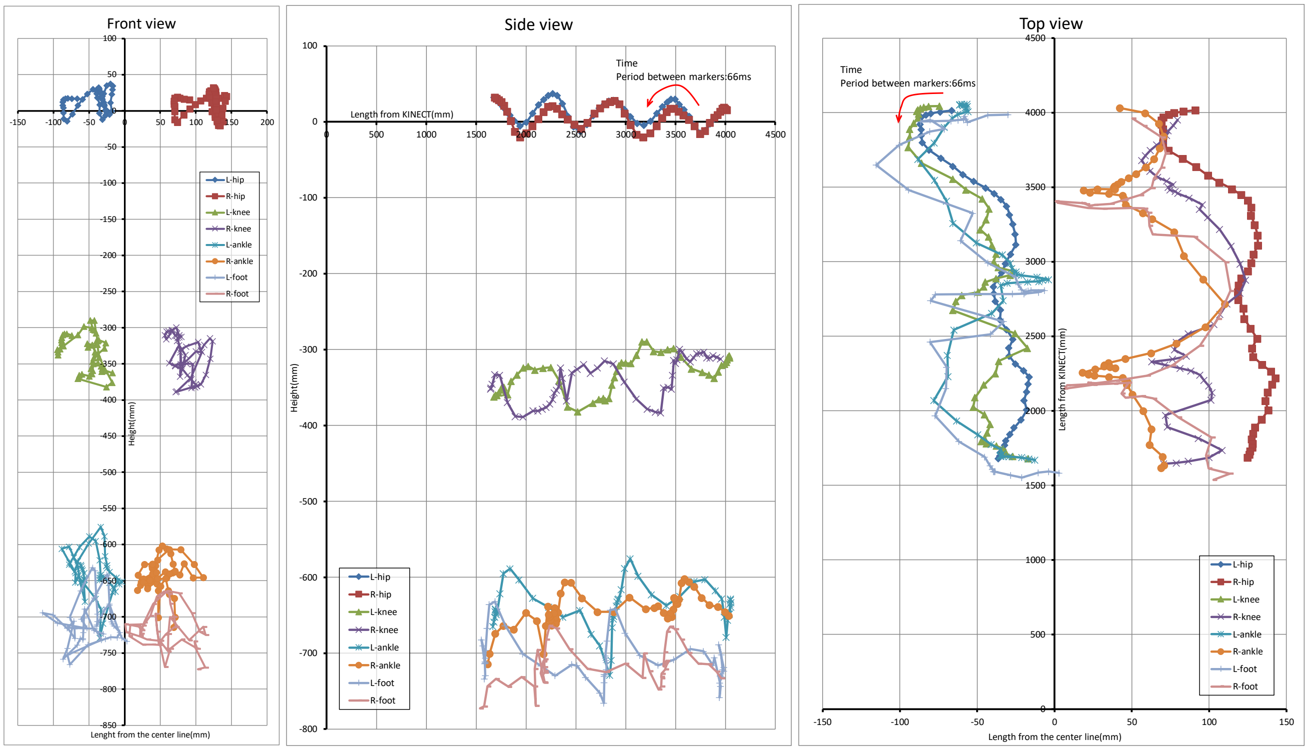
A top view curve in each graph shows the change of length from the center line. From Figs. 14 and 15, measured data were shifted to the right when KINECT was used. However, left and right feet were put on the center line one after the other in the case of on-line walking (Fig. 15 (c)). This means that the measured length from the center line is basically accurate.

When a participant exaggeratedly rotated his/her left-foot away from the center-line in the circumduction gait, curves of his/her left knee, ankle, and foot in the top view clearly showed their motions as shown in Fig. 16 (c). Moreover, their curves in the side view showed that their motions in the vertical direction were bigger than those of the right foot. KINECT detected exaggerated motion with high accuracy.

One physiotherapist said that curves in the side and top views would be useful to instruct hemiplegia patients to walk more generally.

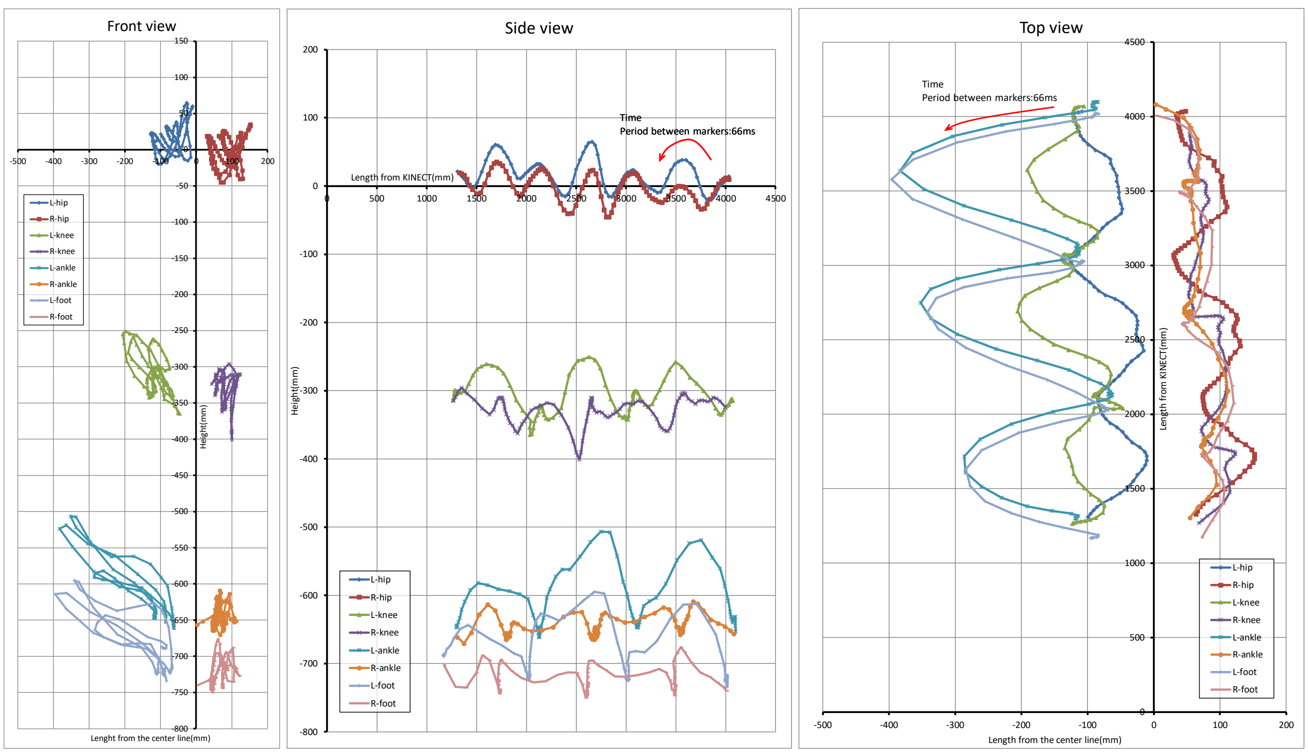
## VII. MEASURING AND ANALYSING A WALKING GAIT FOR WALKING DISABILITIES

We measured angle velocity at the terminal stance to initial swing, angle between a foot and floor at the terminal swing, and average FS using a proposed gait monitoring device and KINECT. Positions of joints on the lower body measured by KINECT were presented with the side and top




(a) Front view (b) Side view (c) Top view

Figure 15. Positions of joints on the lower body for on-line gait



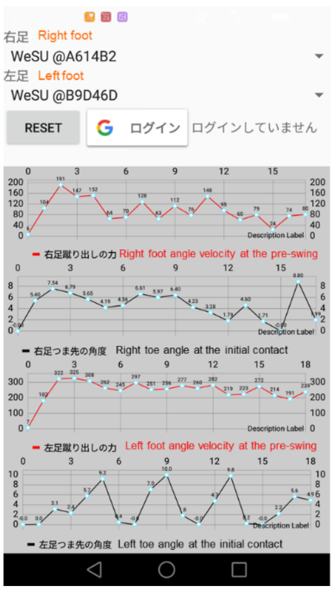

(a) Front view (b) Side view (c) Top view

Figure 16. Positions of joints on the lower body for circumduction gait

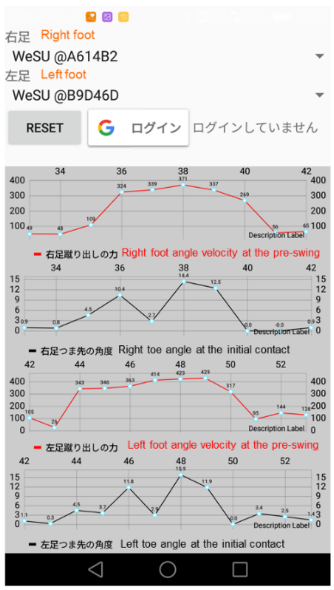
view formats. Since a gait monitoring device did not work well at the first measuring date, we measured four patients using KINECT only at that day (17/08/2017). We measured different four patients using a gait monitoring device at different date (23/08/2017).

## A. *Measured by a walking gait device*

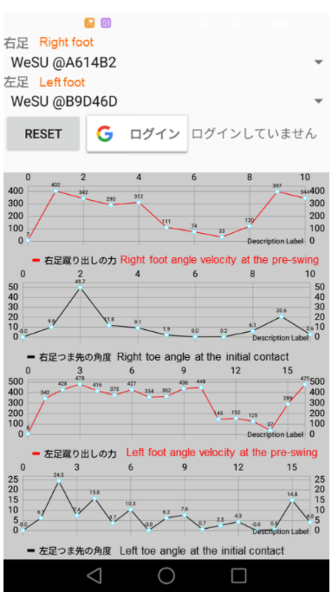
Measured data for four walking disabilities by a walking gait device are showed in Fig. 17. Profiles of patients are listed in Table VII. Since Patient A has paralysis in right-side, his right foot angle velocities at the pre-swing are lower than those of his left foot. His right foot angles at the initial contact are approximately equal to an average foot angle of walking disabilities listed in Table VI. However, those of his left foot are changed so hard. Patient B has paralysis in his left-side. However, his physical strength is weak, and there is not clearly difference between his right and left foot. Patient C and D do not have any paralysis. Their physical strengths are a little weak, and their toe angles at the initial contact are basically low.



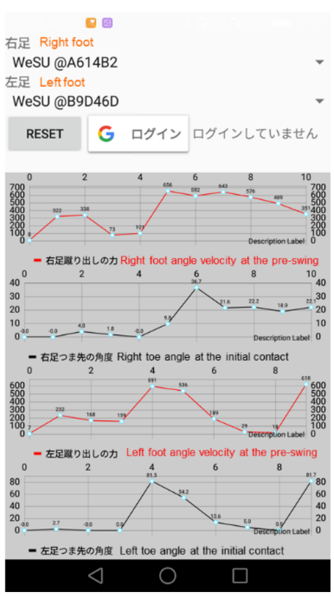

(1) Patient A




(2) Patient B




(3) Patient C




(4) Patient D

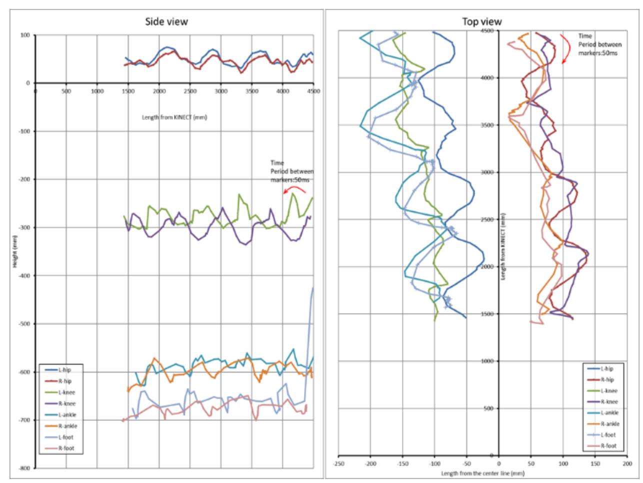
Figure 17. Measured data with a walking gait device

Table VII. Patient profiles measured by a gait monitoring device

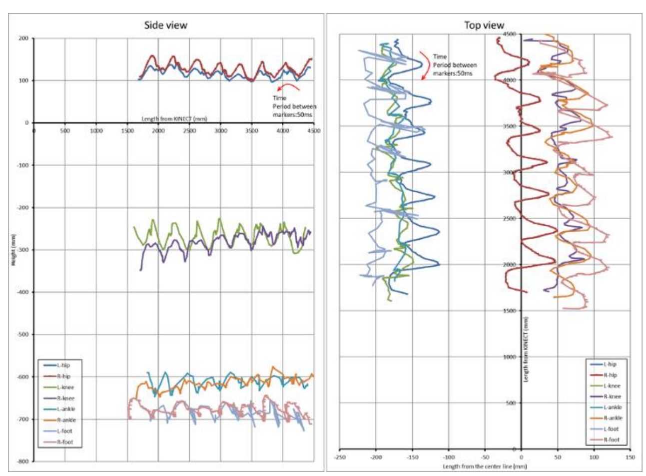
| Patient | Sex | Age | Symtom |
|---|---|---|---|
| A | Male | 77 | Stroke<br>Right-side paralysis |
| B | Male | 67 | Stroke<br>Left-side paralysis |
| C | Male | 80 | Cervical myelopathy<br>Lumgago |
| D | Male | 70 | Quadriplegia<br>Numbness in both shoulders and fingers |

## B. *Measured by KINECT*

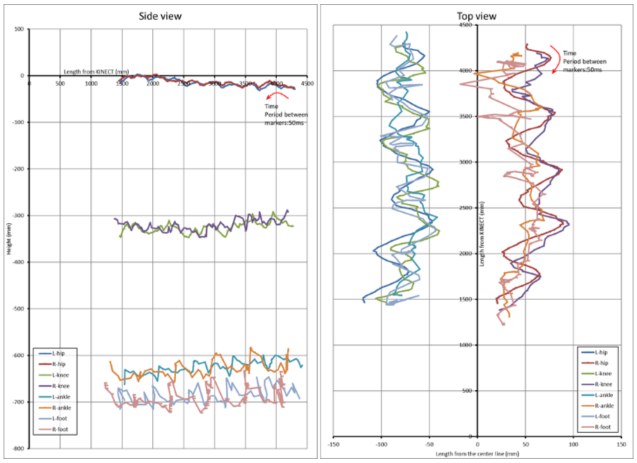
Measured data for four walking disabilities by KINECT are showed with the side and top-view as shown in Fig. 18. Profiles and average FS for patients are listed in Table VIII.



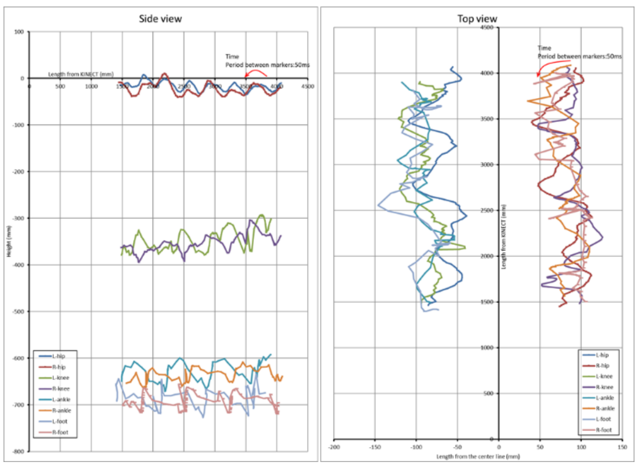

(1) Patient E




(2) Patient F




(3) Patient G




(4) Patient H

Figure 18. Side and top views for joints of walking disabilities

Table VIII. Patient profiles and foot stride (FS) measured by KINECT

| Patient | Sex | Age | Symtom | Av. FS (mm) |
|---|---|---|---|---|
| E | Male | 64 | Stroke<br>Left-side paralysis | 370 |
| F | Male | 79 | Stroke<br>Right lower leg break | 215 |
| G | Female | 64 | Vertebral canal stenosis<br>Both shoulder-ache | 303 |
| H | Male | 73 | Vertebral canal stenosis<br>Left foot downward-pointing toe | 367 |

The average FS for every patient are less than that of walking disability listed in Table V. These data correspond to what physiotherapists said they have a gait disturbance.

The top view curves in Fig. 18 (1) shows that Patient E walks in the circumduction gait for his left-side and has paralysis in his left-side. An average FS of Patient F is very shorter than others. The curve of FS is usually changed symmetry to cross points of both feet. However, his curve is not asymmetry as shown in Fig. 19. The reason is that he uses a cane with his left hand, he sends a cane first, and then his right foot near by the position of cane, and finally his left foot a little beyond his right foot in a gait cycle. Since he leans on a cane with his left hand, his right- hip is shifted to the left.

Side-view curves of Patient G's hip and knee in Fig.18 (3) show that her both hips and knees do not almost move up-and-downward. The chief physiotherapist in this rehabilitation facility says that she must have mastered this walking gait to avoid ache in her both shoulders. There are not any features for Patient H in Fig. 18(4).

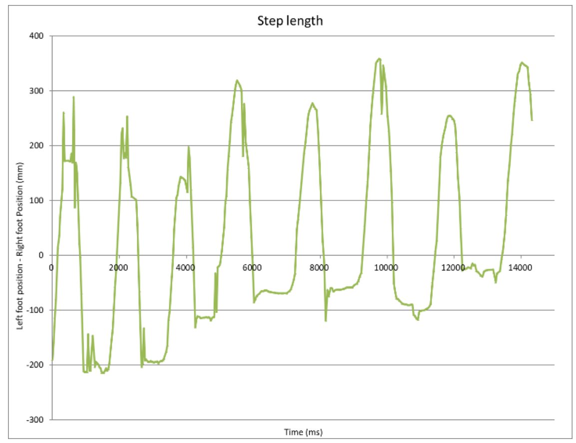



Figure 19. Change of the foot stride of Patient B

## VIII. CONCLUSION

We comparably analyzed walking gait cycle between healthy people and walking disabilities using a wearable device (WD) and KINECT to detect warning signs of falls. On the basis of the results of experiments, we decided to adopt the angle velocity at the terminal stance to initial swing, angle between a foot and floor at the terminal swing, and average FS to detect warning signs of falls. We also developed a gait monitoring device comprised of a smartphone application and a pair of shoes on which WDs were mounted, and proposed using the side and top view formats to present data measured by KINECT. Proposed warning signs calculated for walking disabilities using developed device and KINECT showed clearly difference from healthy people. The proposed presentation format also made clear difference between them. We plan to develop a system that measures effect of rehabilitation quantitatively, and a warning system for fall prevention using proposed gait monitoring device and presentation formats.

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