TABLE 5. TYPES OF RM. AND PERFOMANCE

	I	II	III	IV	V
	%	%	%	%	%
Resist. prfrm. violence	97.7	6.3	78.6	25.9	7.7
Security for patients	100.0	21.9	100.0	37.0	100.0
Avoidance of suicide	100.0	6.3	32.1	18.5	100.0
Sound insulation	100.0	15.6	75.0	11.1	100.0
Isolation (from outside)	100.0	43.8	96.4	29.6	7.7
Isolation (from inside)	32.6	12.5	3.6	0.0	100.0
Medical gas piping	83.7	21.9	10.7	22.2	0.0
Observation camera	100.0	18.8	39.3	25.9	7.7
Observation	97.7	6.3	67.9	7.4	100.0
microphone.	100.0	37.5	100.0	40.7	100.0
Observation window	100.0	100.0	92.9	14.8	0.0
Toilet	67.4	100.0	89.3	0.0	0.0
Toilet door	69.8	100.0	100.0	22.2	0.0
Wash basin	69.8	96.9	32.1	55.6	92.3
Storage furniture	76.7	93.8	35.7	25.9	0.0
Desk					
No. of patients	43	32	28	27	13

TABLE 6. TYPES OF RMS. AND DISEASE

	I	II	III	IV	V	Total
	%	%	%	%	%	
Main disease						
Organic psychosis	19.0	19.0	9.5	47.6	4.8	21
Drug intoxication	18.2	9.1	9.1	63.6	0.0	11
Schizophrenia	37.0	21.9	21.9	9.6	9.6	73
psychopath	40.0	30.0	20.0	5.0	5.0	20
GAF level						
71-80	0.0	50.0	0.0	0.0	50.0	4
61-70	0.0	50.0	0.0	12.5	37.5	8
51-60	6.7	26.7	6.7	33.3	26.7	15
41-50	11.1	22.2	22.2	44.4	0.0	9
31-40	39.4	28.6	27.3	9.1	3.0	33
21-30	35.7	0.0	19.0	11.9	4.8	42
11-20	47.8	0.0	26.1	21.7	4.3	23
1-10	40.0	0.0	20.0	40.0	0.0	5

D. Discussion

1. Necessary ratio of single rooms

Most of psychiatric hospitals ran short of single rooms. A shortage of single rooms occurs troublesome to other patients, lack of privacy and adequate rest. 30% from the patients who stays hospitals less than 3 months needs single rooms. In acute hospital, there requests more number of single rooms.

2. Performance and equipments of ideal single room

Present single room is designed to isolation and observation. For instance, it has a toilet, but without door. It focuses to the most serious patients. But ideal room requests the equipment to improve the daily living such as shower, toilet, desk and so on, to make sure safety such as the function preventing suicide, observation camera, and so on, to provide rest such as sound insulation performance.

Consideration of needed image by types, we get some distinctive feature as follows,

Type I requests the performance against violence behavior and suicide such as locking from outside and isolation.

Type II is comfortable patient's room with adequate rest with toilet, wash basin, furniture, desk and so on.

Type III request the equipments that can lock and be observed from outside, but basic daily living is done in the room. Then it has toilet with door and basin:

Type IV is focuses to Organic Psychosis and Drug Intoxication patients. But they reveal so multi feature that the definition of the room can not be decided.

Type V is the room where the patient locks the door by himself and stays alone with arranging the relationship to other patients. But consideration for suicide and observation is needed.

IV. A STUDY ON ENVIRONMENTAL REMINISCENCE THERAPY FOR MODERATING PERPHERAL SYMPTOMS OF THE DEMENTIA PEOPLE

A. A study of purpose and the proposal of environmental reminiscence therapy

At present, a reminiscence therapy is watched as one of non-drug treatments. When providing this psychological treatment, pictures and small goods are used for this therapy among the patients. Instead of conventional therapy, the authors proposed an ancient tools and architectural space as which or where the dementia people can use and can their own experience directly. The daily living and behavior of dementia people are affected from memorial scene and fondly-remembered architectural space. The authors attention that the concept of "Life Review" will be useful for the therapy.

In present clinical psychological therapy, the evaluation of reminiscence therapy is estimated by the quality and quantity of conversation, and the record and observation of emotion. Our study will propose the therapy, which can promote another efficient from former therapy, and which reminds past days by each other as a one's pace, in the architectural reminiscence space that is introduced into the day care dept. for dementia people. The authors call this therapy as Environmental Reminiscence Therapy.

FIGURE 1. THE MEMORIAL STREET IN SAIHAKU HOSPITAL IN TOTTORI PREF.

FIGURE 2. ENVIRONMETAL REMINISCENCE THRAPY IN THE MEMORIAL STREET

B. Interview Investigation for the People in *Nambucho-Tottori Pref.*

In order to get some evidences to create design concept of the memorial street, we investigated 48 aged people in that village as interview. Average age of the people is 75 years old. Old houses with a straw-thatched roof and old fire place were talked by everybody as a memorial tools and scene. It depends on each region.

C. Design of the Memorial Street

Referring the results of interview, we decided the design of the memorial street. Whole hospital was designed for warm environment with wood and paper which occurs Japanese taste. A lattice door of patients' room, Tatami-mat bench, lamps with traditional paper, a ceiling with bamboo was introduced into the hospital interior. We aimed to make a neighbor become intimate, and the memorial street melt into the hospital design.

Relational rooms for the demented day care are located on the memorial street. A resting room, cooking room, bath and staff rooms except day care room as the reminiscence room are located, and each room are designed what occur the interior. The wall of the street was made with wood wall. We installed the equipments, which stimulate five senses. One of them is the equipment lighting system which direct morning and evening sky by the changing the color of light.

We provided many elements into the many points of the memorial street, and aimed to introduce their own scene that appear their distance childhood memory. At present, red tile roofs, a sign of Hossho-ji station, a poster of kabuki-play at Hossho-ji district and the pictures of Hossho-ji train were installed; in near future we will install some other elements, for instance fire-place into the therapy room, immediately.

D. Evaluation of Environmental Reminiscence Therapy

Dementia disease is consisted with core disease occurred by disorder of brain directly and peripheral disease with psychological, situational and physical factors. This study watches that peripheral disease is affected with environment strongly, tries to verify environment reminiscence therapy through architectural space and circumstances.

As one of therapies of day care program, it was observed that a certain demented patient revealed following change through a tea party with other patients and staffs in the memorial street for one hour. 1) Stability of emotion (conversation with staffs and complete conversation), 2) Easing of mental stress (reducing the stress with long time injection), 3) Lasting concentration (long time conversation), 4) Becoming five senses actively.

TABLE 7. SCORE BY MOSES SCALE

	Female, 89 year old		Female 74 years old		Male 84 years old	
period	before	after	before	after	before	after
Depressed mood	15	8	15	9	11	9
	Recognition for 2 items		Recognition for 3items			
Irritation			15	9	11	12
	No disease		Recognition for 1 item			
confination	17	14	21	20	19	19

In order to get objective evaluation of the change for the demented in reminiscence environment, psychological test was held with dementia scale in the field of clinical psychology. MOSES Scale was adopted to this study. It is the one of the best scale for the demented and used in general field in clinical psychology and evaluated its reliance. This scale is multilateral, and can measure ADL, emotion and the change of peripheral disease of dementia. Observation subjects person are three dementia people. Before test, the occupational therapist measured the degree of dementia by MOSES based on the medical chart. Concrete environmental reminiscence therapy was a kind of tea party with talking about old memories of this district in the memorial street. After one hour therapy, the Occupational Therapist and the nurses evaluated the people by MOSES scale during tea party. Reducing the point from before observation to after one, the items recognized as improvement.

We could get the same result that the past clinical psychological study mentioned that reminiscence therapy is effective for stability of emotion, especially improvement for stress of the demented.

E. Conclusion

A reminiscence therapy is recognized to improve the peripheral disease of the dementia, and environment reminiscence therapy is cleared to verify for many demented people through installing the memorial street into the hospital. It is confirmed by not only observation but also clinical psychological scale.

It is easy to conclude that installing memorial scene like these studies is fake or theme park. However for demented who forgot everything in a few minutes, what is the real? What is the environment the demented recognize? We have to reconsider them.

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Quantitative Evaluation of Movement Using the Timed Up-and-Go Test

Detection of Task Phase and Clinical Application to the Rehabilitation of Hemiplegic Patients

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Many hemiplegic patients who have suffered strokes need rehabilitation. The expected outcome of rehabilitation is the patient's independence and freedom from the aid of a nurse and a consequent improvement in the quality of life. Thus, occupational and physical therapy should be carried out efficiently, and the development of support technology for therapy is important. Developing independence in basic activities is essential in the early stage of rehabilitation as it influences the subsequent recovery of the patient's normal way of life. The timed up-and-go test (TUG-T) is a simple technique for evaluating competence in the following basic activities: standing up from a chair, walking forward, turning around, walking back to the chair, turning one's back to the chair, and sitting down. The total time taken to complete the TUG-T is used to predict the risk of falling [1]–[4]. However, no objective criteria exist for demarcating each activity phase of the TUG-T. At present, these are evaluated subjectively based on the experience of the therapist, and thus it is difficult to obtain objective data for clinical rehabilitation. Conventionally, the therapist evaluates how well the patient performs the TUG-T, at a detailed motion level, and confirms the patient's problems with particular activities. The therapist then determines ways to resolve the problems. Although the TUG-T measurements are easy to perform and accurately predict the risk of falling, it is necessary to perform an evaluation applying appropriate objective data. Acquiring quantitative data would be of great benefit in a rehabilitation program.

Measurements during clinical rehabilitation have been attempted using a triaxial accelerometer to measure the activity objectively, which allows a quantitative evaluation. The triaxial accelerometer enables motion evaluation in the frontal, sagittal, and horizontal planes by measuring the motion in the anteroposterior, vertical, and lateral directions, respectively. The acceleration signals can also be used to evaluate muscle power, joint function, and postural reflexes. Consequently, this assessment provides key information for evaluating the walking activity phase and the other basic activity phases of the TUG-T.

Previously, the activity was examined to evaluate quantitatively using only the signal from an accelerometer attached at the waist because it was believed that it would facilitate measurement during clinical rehabilitation [5]. The literature on evaluating posture using commercial, low-priced, accurate accelerometers is extensive [6], [7].

During clinical rehabilitation, the measurement method should not restrain the subject with too many sensors. Angular measurements can be derived using Kalman filtering of the direct current (dc) element of the acceleration signal. However, this method has a reported error margin of $\pm 2°$ [6]. During clinical rehabilitation, it was difficult to identify hemiplegic walking from the angular displacement signal. It was also difficult to identify the activity phase clearly, which was our objective, using the acceleration signal alone.

A waist gyrosensor is useful for measuring the postural displacement with high accuracy. The posture can be determined by measuring the acceleration and angular velocities, although this method has never been used to evaluate and verify continuous activity from static sitting to walking [8]–[10]. Therefore, an accelerometer and rate gyrosensor was attached to the subject's waist and lower limbs to evaluate postural displacement. A further objective was to identify the activity phases of the TUG-T. Trained therapists measured the time for each activity phase from a videotape recording (VTR) of the TUG-T for reference [11]. This combined accelerometer or gyro method was used during clinical rehabilitation sessions during which the subject performed the TUG-T independently or while being supervised. Under both conditions, the walking phase activities extracted from the TUG-T data were compared qualitatively.

Method

The Measurement System

The measurement system used for the TUG-T consisted of two sensor units (Gyrocube), a multitelemeter system (WEB-5000), and a personal computer with a built-in analog-to-digital converter.

Each sensor unit can measure the three axes of acceleration (the waist accelerometer measures ± 3 g with a sensitivity of 1.33 V/g, whereas the lower limb sensor measures ± 5 g and 0.80 V/g, respectively, with a frequency response dc of 60 Hz) and the three axes of angular velocities (angular velocity ratings $\pm 400°$/s, sensitivity 10.0 mV/°/s, and frequency response

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dc of 40 Hz). The sensor measured $30 \times 40 \times 20$ mm^3 and weighed 7 g.

The signal from the sensor unit was recorded on a personal computer at a sampling frequency of 128 Hz via the multitelemeter system (the high cutoff frequency was 30 Hz).

The transmitter of the multitelemeter system measured $128 \times 80 \times 28$ mm^3 and weighed about 300 g.

Measurement Method

The measurement task was based on the TUG-T introduced by Podsiadlo and Richardson [11], and the procedure is as follows:

1) The subject sits with his or her back in contact with the back of the chair (the seat is 460 mm high and lacks armrests)
2) The TUG-T begins with the therapist's go sign and the subject stands up (standing up)
3) The subject begins walking (walk 1)
4) The subject turns around a post placed 3 m away from the chair (turn 1)
5) The subject walks back toward the chair (walk 2)
6) The subject turns away from the chair to sit down (turn 2)
7) The subject sits on the chair (sitting down).

The acceleration measurement points are at the waist dorsally (near the second lumbar vertebra) and at the lower limb on the side that takes the first step. Figure 1 indicates the positions of the accelerometer and gyroscope. The activity was captured with a CCD camera (EVI-D30) and recorded with a VTR (GV-D900NTSC). The therapist recorded the duration of each activity phase using a stopwatch while watching the video. Ten young, healthy subjects and 20 hemiplegic patients from Fujimoto Hayasuzu Hospital, Japan, were studied (Table 1). Twelve experienced therapists from the clinical rehabilitation center measured the durations of the activity phases. For safety, a therapist stood beside the hemiplegic patients during the activity.

The Method of Detecting and Evaluating the Activity Phase of the TUG-T

The data gathered from the healthy subjects were used to identify the data from the sensor signals corresponding to each activity phase. The features of these corresponding points were examined, and a method of detecting each activity phase was proposed. Figure 2 shows a typical example of the TUG-T in a normal young healthy subject (mean age, 21.0 ± 2), and Figure 3 shows the flowchart used to detect the phase changes. First, each activity phase was identified from the video images using a stopwatch.

From the perspective of rehabilitation training, the ability to stand up and walk is the most important task leading to the activity. Therefore, these two phases must be identified.

1) *Standing-up phase*: During the sitting phase, the waist pitch angular velocity, which is integrated in the angular velocity per unit time, is determined uniquely. In fact, the angular velocity in the sitting phase should be zero. In the healthy volunteers, the waist pitch angular velocity is almost zero, although a maximum waist pitch angular velocity of 10°/s was observed in a healthy subject. Therefore, the standing-up phase occurs when the output of the waist gyrosensor in the pitch direction exceeds a threshold of 10°/s.

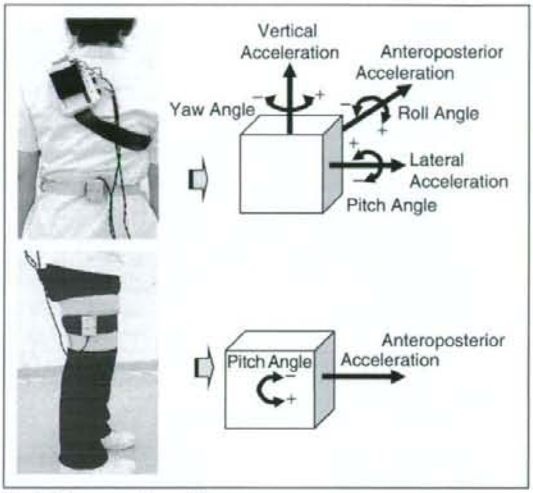

Fig. 1. Sensor unit positions.

Table 1. Subject profiles.

Case	Sex	Age (years)	Paralyzed Side	L/E Br. Stage	Gait Level
1.	Female	66	Right	III	Supervised
2.	Female	51	Right	III	Supervised
3.	Female	82	Right	IV	Supervised
4.	Male	74	Left	IV	Supervised
5.	Male	83	Left	IV	Supervised
6.	Male	65	Right	IV	Supervised
7.	Male	39	Right	III	Supervised
8.	Female	66	Left	IV	Supervised
9.	Female	65	Left	IV	Supervised
10.	Female	62	Right	IV	Supervised
	(Mean age, 65.3 ± 13)				
1.	Female	63	Left	IV	Independent
2.	Female	75	Left	IV	Independent
3.	Male	75	Right	IV	Independent
4.	Female	78	Right	IV	Independent
5.	Male	74	Right	III	Independent
6.	Female	70	Left	V	Independent
7.	Male	57	Right	IV	Independent
8.	Male	70	Right	IV	Independent
9.	Male	74	Left	IV	Independent
10.	Female	76	Left	IV	Independent
	(Mean age, 71.2 ± 6)				

Developing independence in basic activities is essential in the early stage of rehabilitation.

Similarly, the start of the sit-down phase is when the pitch angle is below 10°/s.

2) *Walking phase*: This phase is defined as the first instance when the output of the lower limb gyrosensor in the pitch direction exceeds the threshold Th_w, which equals 10°/s. In fact, the zero-crossing time (the time when the readout is no longer zero) is sufficient to indicate the start of walking. However, to eliminate the effects of the subject swaying and swinging, the value of 10°/s instead of the zero-crossing time was used.
3) *Turning phases*: For turning, the turns while walking (turn 1, t_1) and for sitting down (turn 2, t_2) were identified by a large angular velocity signal in the yaw direction from the waist sensor. To obtain t_1 and t_2, the yaw direction angular velocity signal was processed by applying low-pass filtering using second-order Butterworth filters with a cutoff frequency equal to the walking cadence.

To obtain T_{max}, the time when the yaw direction at the waist is the maximum, and ω_{max}, the maximum value of the yaw direction, β as 35% of ω_{max} was first determined empirically. When the angular velocity exceeded $\omega_{max} \times \beta$, this gave the duration of t_1 (the blue line in Figure 2). Furthermore, the value of $\omega_{max} \times \beta$, which is one step before and after the time of $\omega_{max}(T_{max})$, was considered. When the angular velocity exceeded $\omega_{max} \times \beta$ within $T_{max} \pm \alpha$, these periods (the green line in Figure 2) was added to the duration of t_1 (red line).

The duration of turn 2 (t_2) was obtained in a similar manner. Both made use of the necessary assumptions that turn 1 occurs not long after the start of the walking phase, and turn 2 takes place before the sitting-down phase.

4) *Sitting-down phase*: The sitting-down phase begins at the end of turn 2 and ends when the output of the waist gyrosensor in the pitch direction falls below the threshold Th_s.

When the proposed method was applied to hemiplegic patients, it was taken into account that the movement of both legs during walking is not symmetrical. Therefore, the value of α was set to double the period of one step by a healthy volunteer.

These parameters were uniquely determined from the healthy subjects. The maximum error was included in the proposed assumptions.

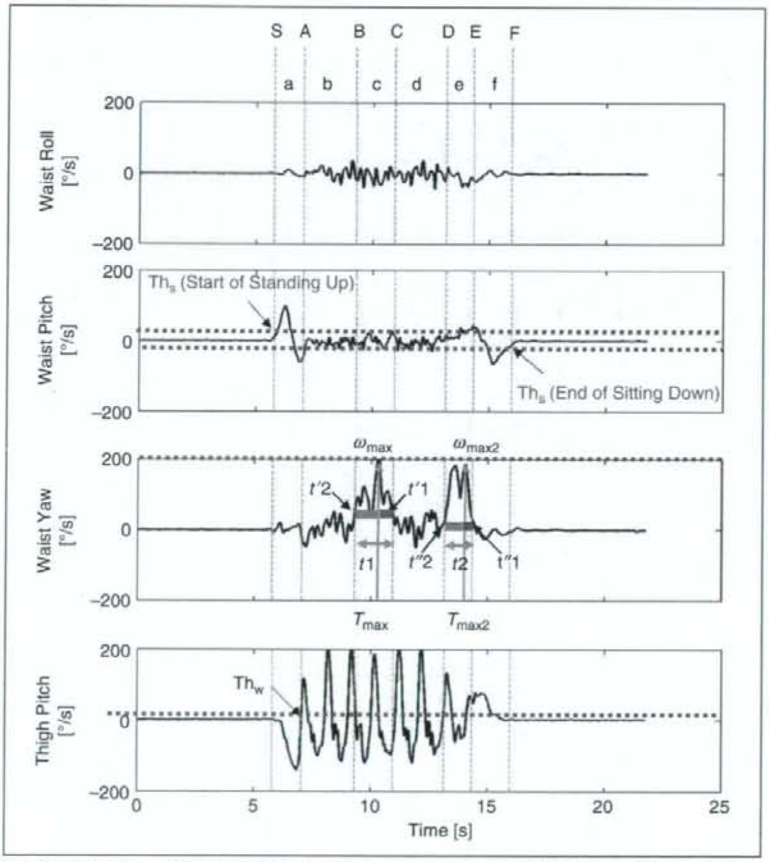

Fig. 2. Typical angular velocities and the points at which the phase changed in a young subject during the TUG-T: (section a) standing up, (section b) walking 1, (section c) turn 1, (section d) walking 2, (section e) turn 2, and (section f) sitting down.

Clinical Application of the Proposed Method

The subjects were 20 hemiplegic patients. Ten subjects were able to walk independently, whereas the other ten could only walk under supervision. Table 1 shows the details of the subjects. The various activity phases were identified, and their durations were obtained from the angular velocities signals during the TUG-T. The data were examined for qualitative differences in walking among subjects, and the independent and supervised walking subjects were compared. The cadence was calculated from the signals for walk 1, turn 1, and walk 2. The root mean square (RMS) value of the acceleration signal, and hence the coefficient of variation (CV), was calculated from the walking cycle. The acceleration signals were compared using these values and the direction of movement. The t-test was used for statistical comparisons, and Bland-Altman plots were used to evaluate the accuracy of our method. Using this method, the data were compared to the measurements made with a stopwatch while the therapist was watching the VTR.

Results

Correlation Between the Proposed Method and Therapists' Measurements

Figure 4 shows that the activity phases of a typical hemiplegic patient identified using the proposed method was similar to the results with a healthy volunteer. The results using the

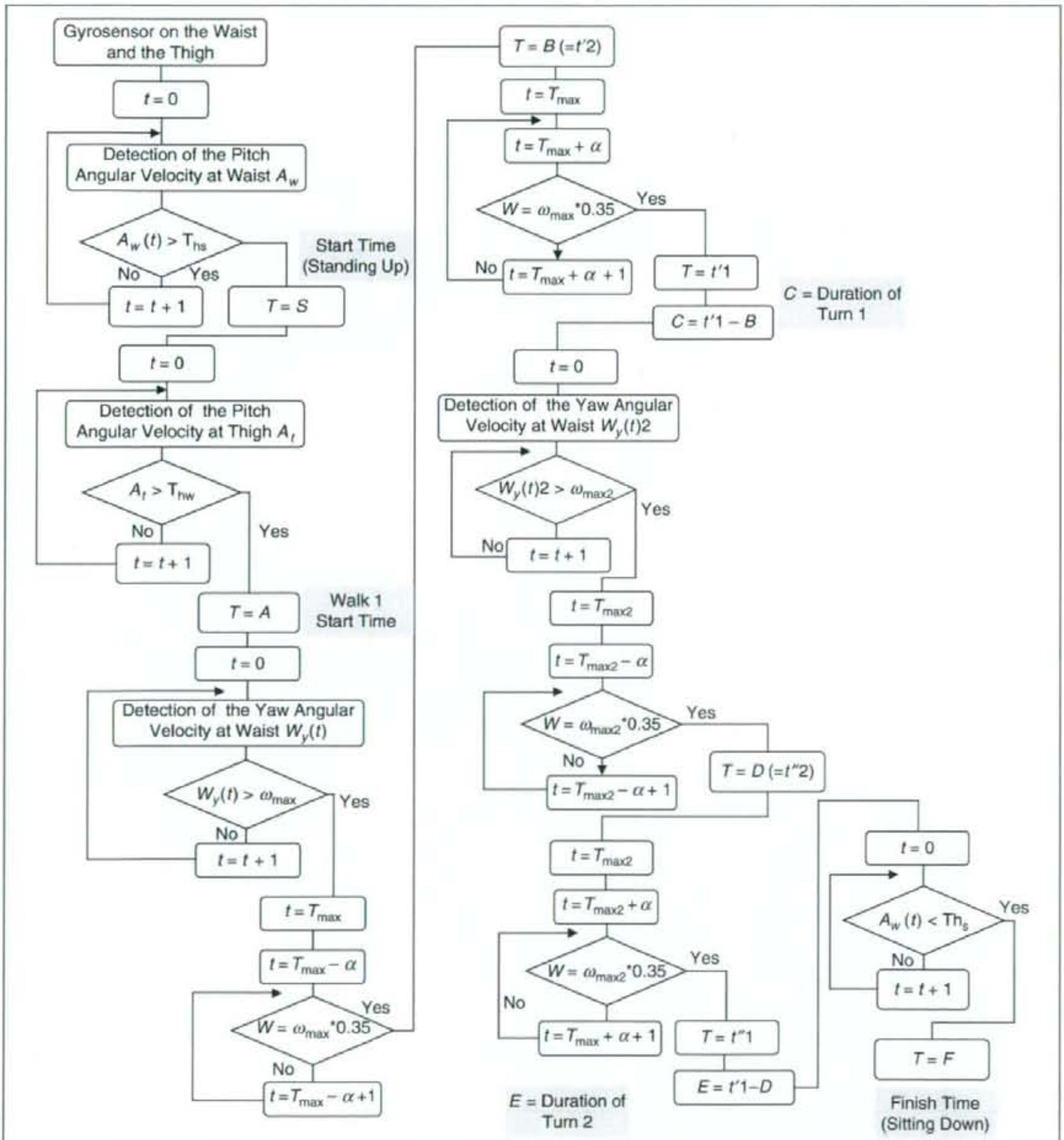

Fig. 3. Flowchart used to detect the changes in phase.

proposed method were strongly correlated with those based on the therapists' observations (Figure 5). Furthermore, most values were included in the Bland-Altman plot within ±1.96 SD (Figure 6).

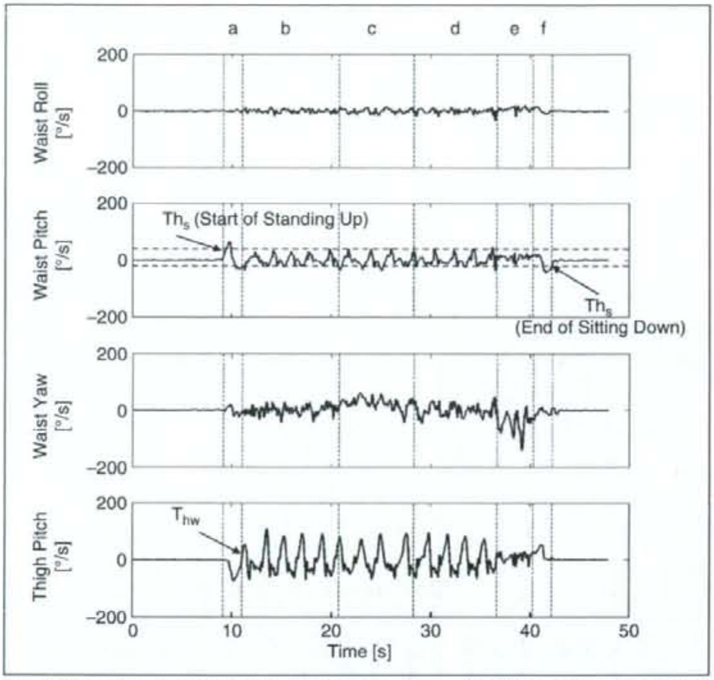

Fig. 4. Typical angular velocities and the points at which the phase changed in a hemiplegic patient during the TUG-T. (a) Standing up. (b) Walking forward. (c) Turn 1. (d) Walking backward. (e) Turn 2. (f) Sitting down.

One feature of the signal for a hemiplegic patient was that the angular velocities in the roll and pitch directions were large, whereas those in the yaw and lower limb pitch directions were small. The results suggested a good correspondence between the times measured by the therapists and the times estimated using our method.

Clinical Application of the Proposed Method

1) Comparing the total TUG-T time, the supervised group took longer than the independent group ($P < 0.05$; Figure 7).
2) Comparing the duration of each activity phase, the supervised group took longer than the independent group for walks 1 and 2 (walk 1: $P < 0.05$, walk 2: $P < 0.01$), while no significant differences were observed for the other activity phases (Figure 8).
3) Comparing the RMS values of acceleration, the supervised group had lower values than the independent group. Specifically, the RMS value in the lateral and vertical directions was smaller ($P < 0.01$; Figure 9).
4) Comparing the CVs, the supervised group had a higher value than the independent group in the lateral direction ($P < 0.01$; Figure 10).

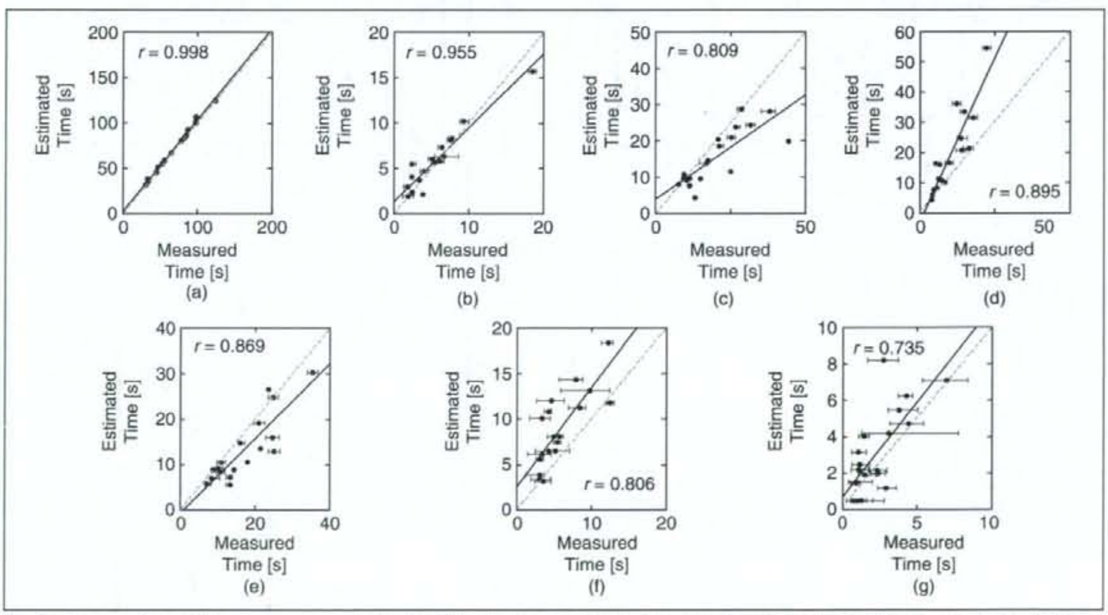

Fig. 5. Comparison of the time measured to hemiplegics by the therapists and that estimated using our method. (a) Total time. (b) Standing up. (c) Walking forward. (d) Turn 1. (e) Walking backward. (f) Turn 2. (g) Sitting down. (The error bar indicates the minimum and maximum values. The solid and broken lines show the regression and identity lines, respectively.)

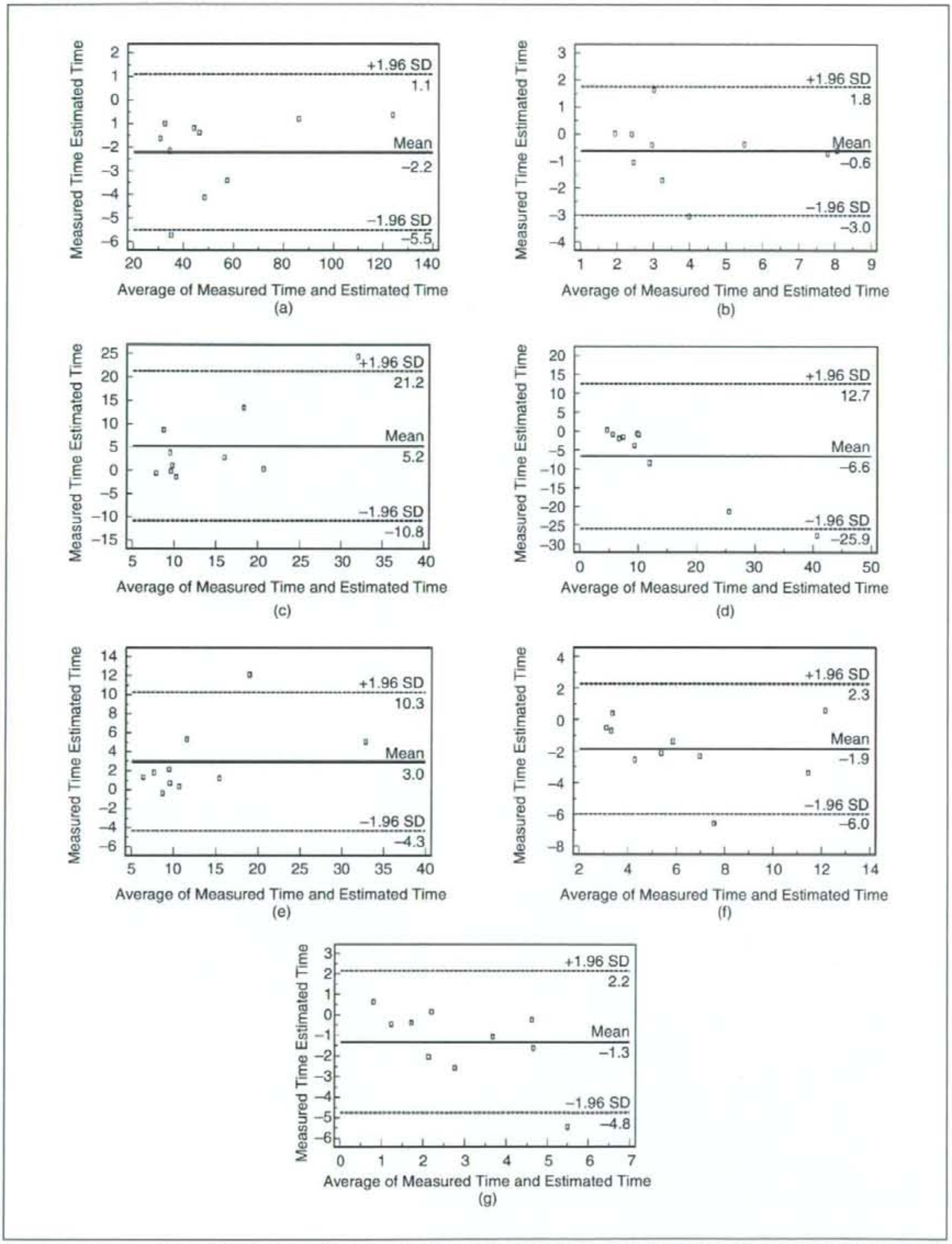

Fig. 6. Comparison of the time measured to hemiplegics by the therapists and that estimated using our method with the Bland-Altman plot. (a) Total time. (b) Standing up. (c) Walking forward. (d) Turn 1. (e) Walking backward. (f) Turn 2. (g) Sitting down.

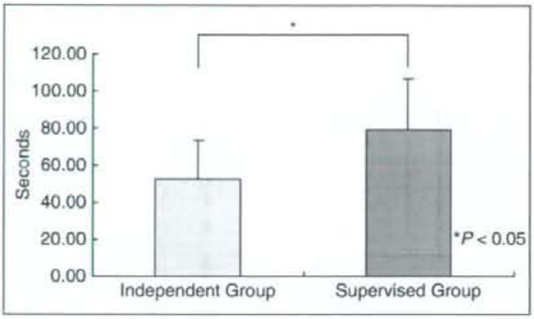

Fig. 7. Comparison of the total times between the independent and supervised groups.

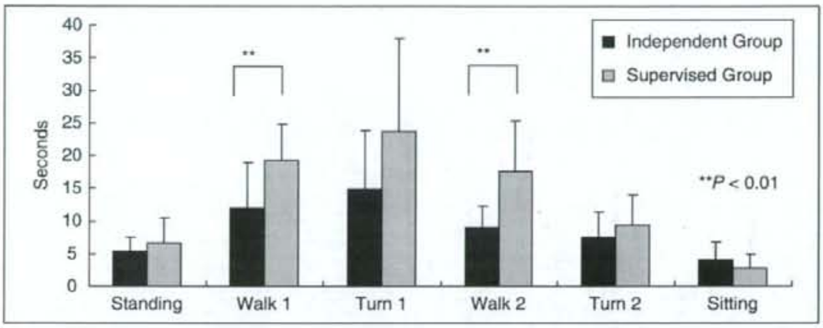

Fig. 8. Comparison of the time for each activity between the independent and supervised groups.

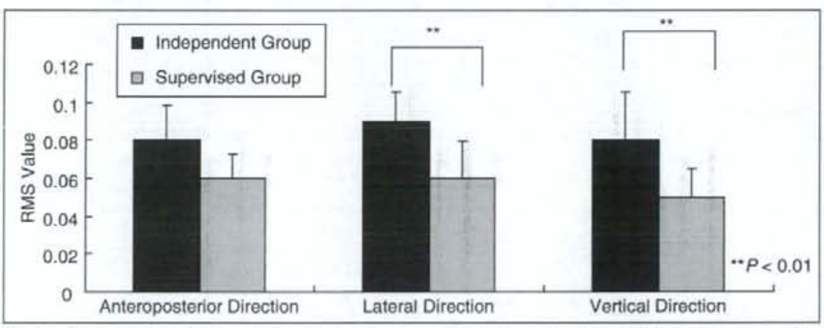

Fig. 9. Comparison of the RMS value for each direction between the independent and supervised groups.

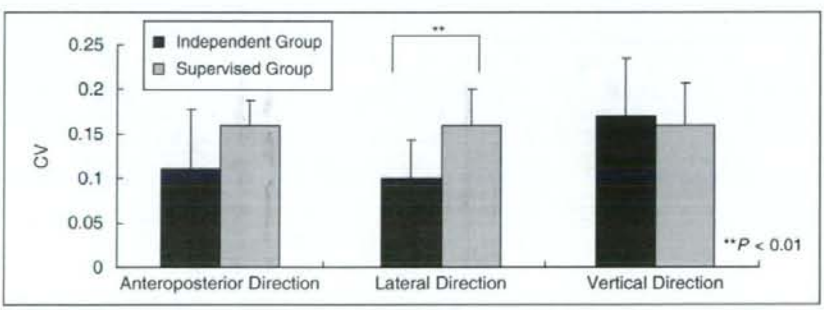

Fig. 10. Comparison of the CV for each direction between the independent and supervised groups.

5) Looking at typical data for supervised and independent subjects, the RMS value for the supervised subjects varied widely at the beginning of walking after standing up (Figures 11 and 12). Similarly, the RMS value varied widely just before sitting.

Discussion

Activity Identification and the Use of Acceleration and Angular Velocity Measurements

The TUG-T is a convenient first test used in teaching fall prevention. The general practice in current research is to measure the activity time in the TUG-T with a stopwatch. However, the TUG-T consists of several activity phases (i.e., standing up, walking 3 m, turning, and sitting down). To better understand the complete performance, it is necessary to evaluate the consecutive sequence of activities. However, because it is difficult to isolate individual problems within each activity phase in the clinical environment, it becomes necessary to identify each activity phase and evaluate each individually.

The Method Used to Detect the Activity Phases in the TUG-T

The start of the standing-up phase could be detected using the waist gyrosensor signal in the pitch direction. Generally, at the time of standing up, a hemiplegic patient shifts his or her trunk by inclining forward markedly, shifting the center of gravity forward to counter the weakness of the lower limbs. The measurement of the pitch angular velocity of the waist was excellent for detecting this forward inclination. The activity of sitting down is similar to, but opposite, that of standing up.

The start of the walking phase immediately after the standing-up phase could be detected from the pitch direction signal of the lower limb angular velocity sensor. Several methods for evaluating walking quantitatively have been used; e.g., electric goniometers, force plates, and impact acceleration [12]–[14]. However, to identify walking as one of a consecutive series of activities, we obtained excellent results from the pitch angular velocity signal of the lower limb.

The identification of the turning phase from the waist yaw direction angular velocity using the proposed method was correlated with the judgment of the therapist. Some features of walking were then considered when hemiplegic patients made turns. For example, hemiplegic patients turn slowly because the radius of gyration is widened to prevent falls. Another feature of walking in hemiplegic patients is the reduced step length to confer stability. This is evident from the small signal of the lower limb pitch direction angular velocities. A small signal for the yaw direction angular velocity showed rotation of the waist, and the signal for the waist roll direction angular velocity had a large amplitude, reflecting the compensational reaction of the waist in the lateral direction while drawing in the lower extremity.

In addition, the signal of the waist pitch direction angular velocity had a large amplitude. This reflects the left-right asymmetry in walking because of the paralysis. Our results showed that the signal information from the gyrosensors worn during the TUG-T can be used to identify the individual activity phases. Our results from the accelerometer can also be used to analyze the activity effectively. In the near future, we should be able to analyze the consecutive activity phases.

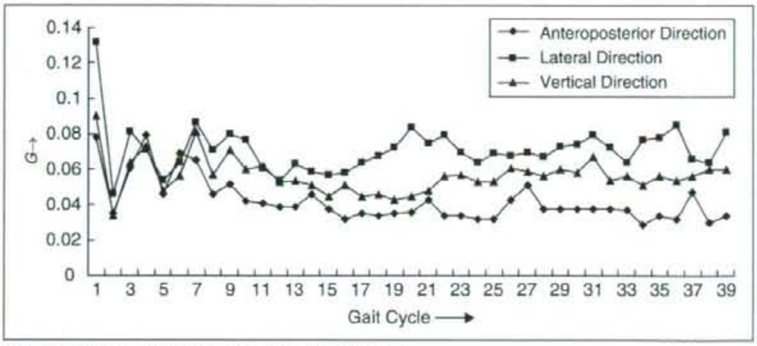

Fig. 11. Typical RMS value in a supervised case.

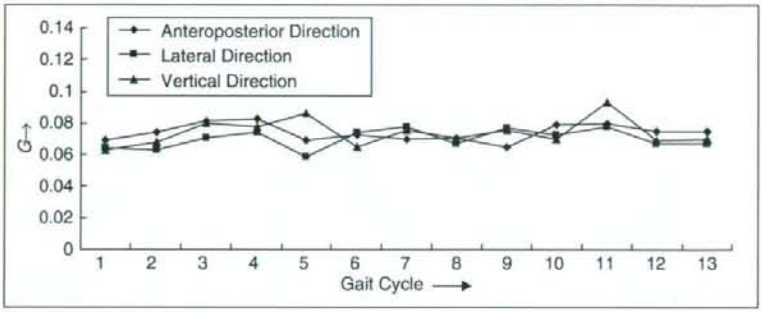

Fig. 12. Typical RMS value in an independent case.

Correlation Between the Proposed Method and Therapists' Measurements

In this study, the length of each activity phase identified using the proposed method was compared with the times measured by the therapist. A high positive correlation was observed between the two, but the error margin for the turn was large among the therapists. Usually, it is easy to identify the activity from a video recording of the frontal and sagittal planes, but to identify the beginning and end of a turn, a video recording of the horizontal plane is necessary. However, it was difficult to cover all angles in the video recording. This explained the comparatively large error margin in the durations of walks 1 and 2, which occur immediately before and after the turn, respectively. The durations of walks 1 and 2 determined using our method tended to be shorter than those determined by the therapists, whereas the duration of the turn between walks 1 and 2 tended to be longer. Therefore, the error margin of both the sensor-derived results and human perception should be considered.

Clinical Application of the Proposed Method

1) Comparing the total TUG-T time, the supervised group took longer than the independent group in walks 1 and 2, but not for the other phases.
2) Comparing the RMS values, the supervised group had lower values than the independent group because the walking velocities of the independent group were greater.
3) The RMS of the acceleration for the supervised group in the lateral direction was smaller because during steady walking, their step length decreases, reducing the vertical movement. Their stride width also decreases, reducing lateral movement. The combination of these two factors prolongs the walking phase.
4) Comparing the CV, the supervised group had a higher value than the independent group in the lateral direction because of less constancy in the stride width.
5) Comparing the data from different subjects, the RMS value varied widely at the beginning of walking after standing up. Similarly, the RMS value varied widely just before sitting because the subject is not steady while walking.

Conclusions

In this study, the combined use of an accelerometer and rate gyrosensor to identify the activity phases of the TUG-T was proposed. For the comparison, trained therapists measured the duration of each activity phase from a video recording. As a result, the proposed identification of the activity phases was well correlated with the therapists' observations. By using both the accelerometer and gyrosensor signals, it was possible to detect the activity phases, which were similar to those observed by the therapists. In addition, the walking activity was extracted from the TUG-T, and the RMS value and CV from the acceleration were calculated in every walking cycle. A qualitative difference between the subjects who could walk independently and those requiring supervision was revealed.

It is currently believed that the TUG-T performance is correlated with the risk of falling. By identifying each activity phase in detail, it is possible to evaluate both the activity from beginning to end (standing up → sitting) and the switches between activity phases (walking → turn → walking). Detailed information was obtained for each activity phase so that the evaluation of the consecutive sequence of activity phases could be realized.

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研究速報

高齢者に対応した歩数カウントアルゴリズムの開発

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Development of a Step Count Algorithm for the Elderly

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Toshiro FUJIMOTO,** Yoshitake OSHIMA,*** Toshikazu SHIGA***

Abstract Step counters are popular for quantifying walking. However, they may not measure the number of steps taken by elderly persons. In this study, we proposed a step count algorithm based on a filter bank and threshold processing to improve the accuracy of an accelerometer-type step counter for elderly persons. The accuracy of the proposed algorithm was compared with the observed steps taken during walking. The waist acceleration signals during self-paced walking were recorded for 74 attendees (age 82.7±8.6 yr, height 148.1±7.6 cm, weight 52.3±7.8 kg) participating in gait training at a rehabilitation services center using a triaxial accelerometer. The participants walked approximately 20 m (10 m in each direction and a turning arc). After seven different band-pass filters were applied to the magnitude of the acceleration vector, a signal related to the step cycle was selected from the outputs of those filters. Then, the number of steps was estimated from this signal using a predetermined threshold. The percent error was calculated as (estimated steps − observed steps)/observed steps × 100. On verifying the algorithm using the data for the 74 elderly subjects, the steps taken by 57 (77.0%) were estimated with less than 10% error. We suggest that our step count algorithm is suitable for estimating the number of steps taken by elderly persons.

Keywords: step counter, acceleration, step count algorithm.

1. は じ め に

今日，日々の身体活動量の低下が血栓症や糖尿病，高血圧症，肥満等のような生活習慣病の危険要因となることはよく知られている．また，身体活動量の増加に伴う効果として，虚血性心疾患のリスクの減少，降圧，糖代謝改善，骨粗しょう症の予防などが報告されている[1-3]．脳卒中片麻痺患者などの身体障害者においても活動量の低下による体力低下が指摘されており，獲得した機能を維持・増進する上で日々の身体活動が重要であるとの報告もある[4]．

歩行は，特別な用具や場所を必要とせず，また身体への無理な負担がなく，安全性にも優れているため身体活動量を増加させる運動として注目されている．厚生労働省は1日の歩数と血圧，HDL コレステロール（善玉コレステロール）との関係を調査している[5]．この国民栄養調査によれば，よく歩いている人ほど血圧は低く，善玉コレステロールが多いという結果が得られている．

歩数を客観的かつ手軽に把握する機器として，歩数計が広く用いられている．近年では，Micro Electro Mechanical Systems (MEMS) 技術の向上とともに多軸の感度をもつ加速度センサを内蔵した歩数計も市販されている．しかし，このような歩数計をもってしても，筋力の低下にともない歩行速度が減少した高齢者や，片麻痺などの運動障害をもち歩行リズムの不規則な高齢者の歩数をカウントすると，正しい値が得られないことが見受けられる[6-8]．Cyarto らの研究結果によれば，Yamax 歩数計を用い老人福祉センターを利用する高齢者を対象に通常歩行時の歩数をカウントしたところ，55%の誤差が観測されている[6]．この問題に対して，これまでの先行研究では得られた加速度を4秒間ごとにフーリエ変換し，パワーが最大となる周

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波数からその区間の歩数を推定する歩数カウントアルゴリズムを提案し，49名中40名（81.6%）の高齢者に対して誤差±10%未満で歩数をカウントしている[9]．しかしながら，この手法は加速度信号を窓（4秒間）ごとにフーリエ変換する必要があり，窓内で周期が変化した場合に歩数カウントの誤差の原因となる．

そこで，本研究では先行研究の経験をもとに，フィルタバンクと閾値処理を用いて加速度波形の変化を考慮した歩数カウントアルゴリズムを提案し，本アルゴリズムの高齢者への対応についてオフラインによりコンピュータ上で検証した．

2. 実 験 方 法

2·1 歩数カウントアルゴリズム

提案する歩数カウントアルゴリズムは，3軸加速度信号の合成，7つの周波数帯域にフィルタリング，適切なフィルタ出力の選択，閾値処理による歩数のカウントという要素を含む．本アルゴリズムでは，まず測定により得られた3軸方向の加速度から加速度ベクトルの大きさを算出する．これはベクトルの大きさは取り付け角度に依存しないため，歩数計の向きにある程度の自由度をもたせることができると期待したためである．次に，加速度ベクトルの大きさをフィルタバンクにより7つの周波数帯域に分離する．各バンドパスフィルタの周波数帯域は0.5～1.0 Hz，1.0～1.5 Hz，1.5～2.0 Hz，2.0～2.5 Hz，さらに各周波数帯域の間の0.75～1.25 Hz，1.25～1.75 Hz，1.75～2.25 Hzに設定し，各フィルタからの出力は概ね単一の周波数成分のみをもつ波形となるようにした．なお，各バンドパスフィルタは1次のバターワースフィルタとした．これら7つのフィルタ出力を時刻毎に比較し，歩数カウントに最適なものを随時選択する．フィルタ出力の比較は，それぞれのフィルタ出力を全波整流し，カットオフ周波数0.1 Hzの1次のバターワース型ローパスフィルタを通した波形で行い，振幅が最大となるものを選択した．この選択された歩数カウント用波形が閾値以上となる点を1歩としてカウントした．なお，歩数カウントの際の閾値は予備実験の結果を踏まえ0.01 gとした．**図1**に，模擬波形に対するフィルタ出力とフィルタ出力の選択，歩数カウント用波形を示す．ただし，図1は3つのバンドパスフィルタ（フィルタ帯域1.0～2.0 Hz，2.0～3.0 Hz，3.0～4.0 Hz）を用いた場合の例である．まず，加速度ベクトルの大きさを模擬した波形を図1(a)に示す．この模擬波形に対するフィルタ出力はそれぞれ図1(b)のようになる．フィルタ出力の大きさを比較するためそれぞれの信号を全波整流し，ローパスフィルタを適用し平滑化した結果，図1(c)のような波形が得られ，振幅が最大となるバンドパスフィルタを選択すると図1(d)となる．図1(c)では，模擬波形の周期にあわせて，0~2秒までは1.0～2.0 Hz，2~4秒までは2.0～3.0 Hz，4~6秒までは3.0~4.0 Hzの周波数帯域を選択している．最終的に，各時刻において選択されたバンドパスフィルタの出力波形をつなぎ合わせた歩数カウント用波形は図1(e)のようになる．この歩数カウントアルゴリズムのブロックダイアグラムを**図2**に示す．今回の実験では実際には回路は作成せず，MATLAB（version7.0, Math Works, MA）によりシミュレーションおよび解析を行った．

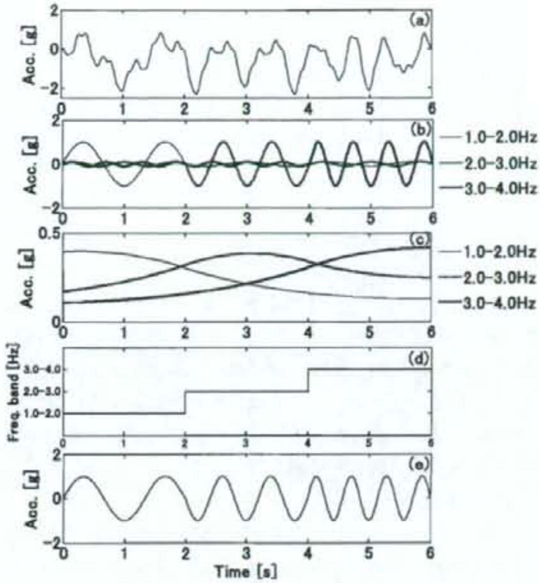


図1　フィルタリング課程：模擬信号（a），バンドパスフィルタリング後の信号(b)，ローパスフィルタ後の信号(c)，選択した周波数帯域（d），歩数カウントに用いる波形(e)

Fig. 1　Filtering process: a dummy signal (a), the band-pass filtered signals (b), the low-pass filtered signals (c), the frequency band of selected filter (d), the synthesized signal from the filtering signals for step count (e).

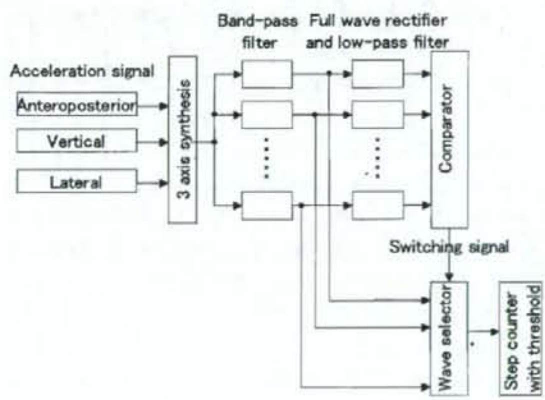


図2　歩数カウントアルゴリズムのブロックダイアグラム

Fig. 2　Block diagram of the step count algorithm.

2・2 測定装置

歩行中に身体に生じる加速度を測定するため，3軸加速度センサユニット（測定範囲±2 g，寸法 30×40×20 mm，重さ 20 g）とマルチテレメータシステム（WEB-5000，日本光電）からなる測定システムを構築し，A/D 変換ボードを介し PC に接続した．なお，センサユニットは，半導体ピエゾ抵抗型3軸加速度センサ（曙ブレーキ，測定範囲±2 g，検出感度 5 mV/g（x 軸，y 軸），4 mV/g（z 軸））と増幅器，カットオフ周波数 50 Hz の1次ローパスフィルタで構成される．A/D 変換する際のサンプリング周波数は 128 Hz，分解能は 12 bit とした．また，参考として市販の加速度歩数計（HJ-720IT，オムロンヘルスケア，以下歩数計）を使用した．なお，この歩数計の取扱説明書には，不規則な歩行や極端にゆっくり歩いたときは正確にカウントできないことがあると明記されている．

2・3 測定方法

被験者は，転倒予防教室に来場している高齢者 74 名（年齢 82.7±8.6 歳，身長 148.1±7.6 cm，体重 52.3±7.8 kg，歩行自立度　自立：52 名　監視：20 名　介助：2 名，歩行条件　杖歩行者：23 名　両松葉杖：1 名　全盲：1 名）とした．被験者には，脳梗塞による片麻痺患者，パーキンソン病患者，膝関節置換術や骨折経験者などの下肢に何らかの機能障害を持つ被験者が含まれた．また，疾病や障害による明らかな下肢機能障害を持たない被験者であっても，多くの被験者は歩行速度が遅く歩行が不安定な高齢者特有の歩容であった．なお，本研究は八日会藤元早鈴病院および千葉大学の倫理委員会の承認を得た後，被験者に実験の詳細を説明し書面にて同意を得て行った．

被験者は，伸縮性のあるベルトを用いて加速度センサユニットを腰背部中央に装着した．歩数計も同ベルトを用いて加速度センサの近傍，約 5 cm の部位に取り付けた．歩行は，屋内廊下直線 10 m 程度を個人に適した速度で往復するものとした．なお，靴等の履物や歩行補助具の指定は特に行わなかった．被験者の安全の確保と歩数の真値（実歩数）を把握するために，理学療法士が被験者の後方を付いて歩き，手持式数取器にて歩数をカウントした．提案した歩数カウントアルゴリズムならびに加速度歩数計の評価は，Cyarto らと同様に，誤差(%)＝(カウント数－実歩数)/実歩数×100 を用いて行った[8]．

3. 結　　果

測定・解析結果の典型例を図3に示す．上から順に前後(a)，左右(b)，上下(c)の加速度原波形，加速度ベクトルの大きさ(d)，本アルゴリズムにより選択されたバンドパスフィルタ(e)，フィルタ処理後の歩数カウント用波形(f)となっている．加速度原波形や加速度ベクトルの大きさにおいて，1歩を示す波形パターンが確認されたが，1歩ごとのピーク値の大きさなどに変動がみられた．一方，歩数カウント用波形は，各時刻においてバンドパスフィルタを選択することで単一の周波数成分のみをもつパターンとなった．また，この波形の周期は，原波形の基本周期（一歩周期）に一致していた．

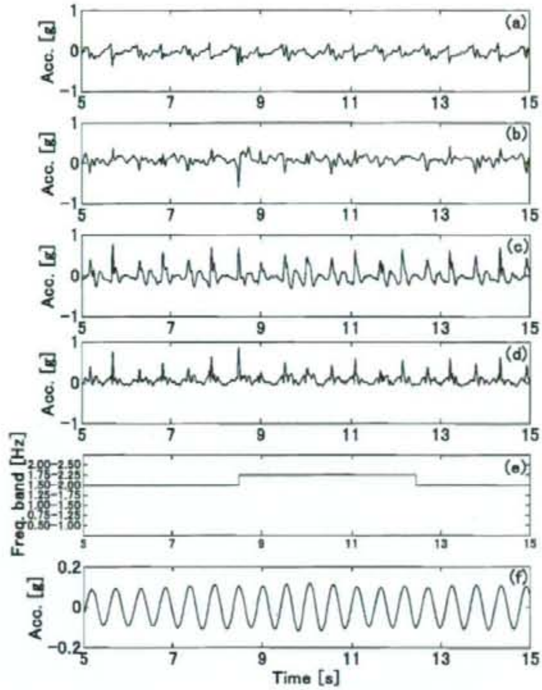


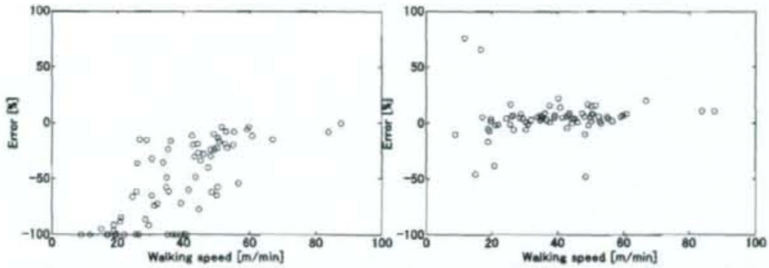
図3　前後方向（a），左右方向（b），上下方向（c）の加速度波形，加速度ベクトルの大きさ（d），選択されたフィルタの周波数帯域（e），および歩数カウントのための波形（f）の典型例

Fig. 3　A typical example of the acceleration signals at the anteroposterior (a), lateral (b) and vertical direction (c), the magnitude of the acceleration vector (d), the frequency band of the selected filter (e), and the synthesized signal from the band-pass filtered signals (f).

図4に市販の歩数計および提案した歩数カウントアルゴリズムによってカウントした測定誤差と歩行速度の関係を示す．本アルゴリズムでは1例を除き約 20 m/min 以下で誤差が増加した．一方，参考のため取り付けた歩数計では約 60 m/min 以下で誤差が増加した．図5に歩数カウントアルゴリズムでの測定誤差のヒストグラムを示す．アルゴリズムの検証実験に参加した被験者74名に対して，本アルゴリズムにより歩数のカウントを行った結果，誤差が±10%以内であったものは 57 名（77.0%）であった．

4. 考　　察

健常成人を対象とした個人に適した速度に対する市販の歩数計の測定精度に関する近年の研究では，平均歩行速度が約 96 m/min で測定精度は 3~37%であった[10]．しか



(a) Step counter (b) Step count algorithm

図4 歩行速度と測定誤差との関係
Fig. 4 Relationship between walking speed and the error of step count.

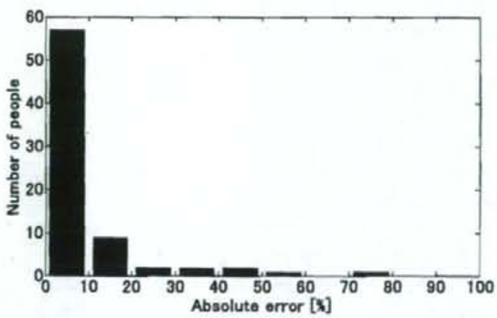


図5 測定誤差のヒストグラム
Fig. 5 Histograms of the error of step count.

し，歩行速度が54 m/min以下の場合，多くの歩数計で誤差が大幅に増加することが報告されている[11-13]．今回の被験者の歩行速度は39.0±15.4 m/minであり，市販の歩数計で大きな測定誤差が出たことは上述の先行研究の結果と一致した．歩行が不規則であり歩行速度が遅いため使用した市販の歩数計の本来の使用条件外であると考えられるが，参考までにこの歩数計で誤差が±10%以内であった被験者は全体の9%であり，測定不能（カウント0歩）となったものは23%であった．今回の結果からも，高齢者，特に下肢機能障害をもつ高齢者に対応した歩数計を考える場合，歩行速度の低下と不規則な歩行を考慮する必要があると示唆された．一方，提案したフィルタバンクと閾値処理による歩数カウントアルゴリズムを適用した場合，誤差が±10%以内あったものは全体の77.0%であった．これは先行研究で得られたフーリエ変換による手法の精度（81.6%）に近い結果である[9]．また，図4に示した通り歩行速度が約20 m/min程度までは高精度で歩数のカウントが可能であり，本アルゴリズムを用いることで高齢者に対する歩数のカウント精度が大きく改善されたと示唆された．

今回用いたアルゴリズムは，7つのバンドパスフィルタにより一歩周期に最も寄与する周波数帯域の信号を選択的に抽出することで，歩数をカウントするための波形が単一の周波数成分になるように設計した．そのため，各フィルタの周波数帯域は線形スケールで等間隔とした．一方で，通過させる一歩周期の幅に着目し，周波数帯域を対数スケールで設定するフィルタバンクも考えられる．そこで，0.50 Hz～2.50 Hzを対数スケールで等間隔で区切った場合の検討を行った．その際のフィルタ帯域は0.50～0.75 Hz，0.75～1.12 Hz，1.12～1.67 Hz，1.67～2.50 Hz，さらにその中間の0.61～0.91 Hz，0.91～1.37 Hz，1.37～2.04 Hzとなる．この周波数帯域で本アルゴリズムを適用すると誤差が±10%以内となる被験者は全体の71.6%であった．このフィルタバンクの場合，高い周波数帯域では帯域幅が低周波帯域に比べて広くなっている．このため，低い周波数帯域つまり歩行周期が長い場合では歩数カウント誤差は小さくなるが，高周波帯域では波形パターンが複雑になり閾値処理ではカウント誤差が大きくなってしまうと考えらる．そこで，本アルゴリズムでは周波数帯域を線形スケールで等間隔に区切り，歩数をカウントするための波形を単一の周波数成分になるように設計し，閾値処理の際の誤差を防止した．

本研究では，歩行のみを動作対象とし実験を行った．本アルゴリズムは，歩行中の加速度振幅が非常に小さい高齢者の歩数をカウントできることから，立ち止まっているときの体の揺れや，電車の揺れなどで誤カウントをする可能性がある．今後は様々な動作を含む日常生活において正確な歩数がカウント可能であるかの検証をしていく必要がある．

5. まとめ

本研究では，フィルタバンクと閾値処理による歩数カウントアルゴリズムを提案し，高齢者を対象に検証実験を行った．今回の実験では，加速度センサを腰背部中央に装着する理想的な測定条件であるものの，74名中57名(77.0%)の被験者に対して誤差±10%以内で歩数をカウントすることが可能であり，本アルゴリズムは筋機能の低下や下肢運動障害をもつ多くの高齢者の歩行に対応できると示唆され

た．

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転倒エアバッグのための転倒検出方法の検討

Examination of the fall detection method for a fall air bag

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Abstract

In order to prevent from an external injury in case of the fall for the aged people, a hip protector is commonly used. However, it is difficult to attach it to the body for a long time, because of difficulty to wear. In order to solve these problems, we are developing the system which prevents externally caused injury by expanding an air bag at the time of a fall. It is necessary to operate an air bag by detecting a fall before the impulse of a fall occurs, in order to develop this system. In this study, we reviewed the algorithm of the fall process, and determined the inflating trigger signals from the algorithm. To determine the triggering signal for the air bag, the mimicking fall, walking and jogging has been performed. The accelerometer was attached to the 16 younger healthy subjects, and the 48 mimicking falls were carried out. This experiment was approved by the ethic committee of Faculty of Engineering, Chiba University and written informed consent was obtained by each subject. The free fall acceleration could be observed around 100 to 300 [ms] before falling down completely. However, misdetection occurred while using the detection algorithm in jogging, which used the accelerometer.

Key Words

fall, airbag, accelerometer, elderly

1．はじめに

近年の食生活や生活環境の変化，医療技術の進歩により日本人の平均寿命は増加し，2025 年には日本の人口の 30%弱が高齢者になると予想されている．[1]このような状況の中，今後は高齢者が長生きするだけでなく，健康に長寿を全うする社会を実現するために，高齢者の QOL を向上させることが重要な課題である．高齢者の QOL を向上させる一つの方策として，「ねたきり」の防止があげられる．「ねたきり」の原因としては転倒や転落による外傷や骨折などが主要な原因の一つであるとの報告がされている．[2-4]さらに，転倒を経験することにより，転倒に恐怖感を抱き，身体活動が消極的になることで，「ねたきり」につながる「転倒後症候群」の問題も指摘されている．[5,6]このことより，転倒時に発生する骨折や外傷を防止することは「ねたきり」を減少させ，高齢者の QOL を向上させる有効な手段であると考えられる．

従来，転倒による事故を防止する機器として，センサやカメラなどを用いて，転倒の危険性が高いベッドからの離床の行動を事前に検出し，看護師や介助者に通報することで，転倒を防止するものが市販されている．[7]これらの機器として，マット型のセンサを用いる方法や，[8,9]画像センサを用いる方

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法[10]などがある．マット型のセンサはベッドの中やベッドの下に設置し，対象者が離床したことをナースコールで知らせるものである．これらのセンサは就寝中の使用しかできず，その他の日常生活中に使用するには有効ではない．画像センサを用いる方法は，CCD カメラの差分情報のみを用い動作を判断することで，プライバシーの問題を解決しつつ，広範囲の行動を検出可能である．しかし，センサを居室内の天井に設置する必要があり，外出時の検出は困難である．

上述したように，現在市販されている転倒防止機器は完全に転倒を防止することは困難である．このため万が一，転倒が発生した場合に速やかに対処可能な機器が必要である．

転倒発生時に外傷を防止するものとしては，ヒッププロテクタが市販されている．これらの装具を用いて行った研究[11,12]では，常時装着することにより転倒による外傷を防止する効果があることが報告されている．しかし，これらの装具は，ずれないようにきつく装着する必要があり，高齢者が装着するには煩わしさがある．また，大腿部頚部骨折を防止する目的で開発されているため，その他の部位の外傷を防止することは困難である．これらの問題点を改善するには，装着が容易で，転倒時に確実に衝撃を吸収可能な機器が必要である．我々は転倒時にエアバッグを膨張させることで，装着が容易で，確実に衝撃吸収が可能な「転倒エアバッグ」の開発をおこなっている．「転倒エアバッグ」を実現するには，転倒を事前に検出し，転倒の衝撃が発生するまでに，エアバッグを膨張させる必要がある．

従来転倒を検出する研究としては，加速度センサや角速度センサを用いて転倒を検出する研究が行われている．[13-17] われわれはこれまで，転倒の発生原因を明らかにし，転倒防止に役立てる目的で，転倒モニタの開発を行っている．[18]開発した転倒モニタは3軸の加速度センサにより転倒前後の姿勢変化と，転倒による衝撃が発生した場合の転倒前後の加速度波形を記録する構成となっている．また清水らは，加速度センサを用いて鉛直方向の変位を検出することで転倒を検出し，外部へ通報するシステム[19]を開発している．しかし，これらの研究は，転倒の衝撃が発生する前に転倒の予兆を検出することはできない．

転倒の予兆を検出する方法としては，足裏の加重センサを用いる方法[20]や，光学式のセンサを用いる方法[21]がある．加重センサを用いる方法は，足裏に加重センサを複数装着し，加重の移動パターンから転倒の予兆を予測するものである．この方法は，身体の姿勢情報がないため，転倒と日常行動の弁別が困難であるという問題がある．また，光学式センサを用いる方法は，身体の腰部と背中の2カ所に光学式の距離センサを装着し，床面との距離の差を計測することで転倒の予兆を検出する方法である．この方法は前後，左右の4方向にセンサを装着する必要があるため，被験者の負担になる．また，屋内では壁や柱が存在するため，常に床面からの距離を測定するのは困難である．本研究では，身体の姿勢や運動の状態を計測可能な加速度センサを用いて転倒の予兆の検出を試みた．本論文では装置の概要と転倒を事前に検出する方法について検討し報告する．

2. 加速度を用いた転倒の計測とアルゴリズムの決定

2.1 転倒エアバッグの概要

開発を行っている転倒エアバッグの概要について述べる．本研究では，転倒時に骨折の危険性が高い大腿部や，後頭部を防御するエアバッグを開発することを最終目標としている．まず，市販の部品を用いた機能試作を行った．Fig.1 にエアバッグの概念図を示す．被験者が転倒を開始した際のトリガ信号からエアバッグの膨張を行う構成とした．Fig.2 に転倒エアバッグのブロック図を，Fig.3 に試作したエアバッグの外観を示す．試作した装置はセンサ部，ガスボンベ，レギュレータ，電磁弁，エアバッグより構成されている．信号処理には 16 ビット CPU(H8 3048 Renesas Technology)を用いた．加速度センサの出力を CPU 内蔵の 10 bit A/D 変換器でデジタルデータに変換し，後に述べる転倒検出アルゴリズムを用い，電磁弁にトリガを出力する構成とした．ガスボンベにはガスによる火災を防止するため，CO_2 ガス(グリーンガス，サンプロ)を用いた．ガスボンベの圧力は 6 MPa であり，そのままでは圧力が高すぎ危険なため，レギュレータ(可変レギュレータ，サンプロ)により 1 MPa に減圧される．減圧されたガスは電磁弁(VZ312-9HS-M5, SMC)によりトリガ信号が発生した場合にエアバッグにガスを供給する．エアバッグには市販の非観血血圧測定用のカフ(YP-914P, 日本光電)200×500mm を用いた．装置の重量はガスボンベの重量を含み 650g である．今後実用化するにあたり，小型軽量で 1 週間程度連続使用する必要があるため，低消費電力の CPU を用いる必要がある．このため処理速度やメモリの制限がある．一方で，転倒検出を行うにはリアルタイムに信号を処理し，転倒と判断した場合にトリガを速やかに出力する必要がある．このため，転倒検出のアルゴリズムをなるべく簡便にし，かつ確実に検出可能とする必要がある．これらの条件を踏まえ，転倒検出のアルゴリズムを検討した．

2.2 転倒検出アルゴリズム

本研究では，転倒の検出にピエゾ抵抗型の 3 軸加速度センサを用いた．ピエゾ抵抗型の加速度センサはセンサ内の錘を支える梁のひずみ変化を抵抗変化として検出するため，加速度センサの出力は重力加速度に相当する直流成分を出力可能である．また，センサが傾斜した際には傾斜角度に従って直流成分も変化するため，3 軸加速度センサを身体に装着した際には姿勢の変化により直流成分が変化する．3軸の加速度の検出方向は身体が直立した状態で前額面に対して垂直な軸を前後方向，矢状面に対して垂直な方向を左右方向，水平面に垂直な方向を上下方向とした．通常の日常生活中は足裏や体の一部が床に接地しているため，身体には鉛直方向に重力加速度が加わっている．従って，被験者に前後，上下，左右の 3 軸方向が計測可能な加速度センサを装着した場合，加速度センサは上下方向に重力加速度を検

Fig.1 Schematic diagram of inflating the air bag