2）機構

装置は前後に移動するテーブル、テーブルを駆動するACサーボシステムならびに制御用ＰＣから構成される。

テーブルの前後方向の駆動にはボールねじ機構を用いて、ボールねじ機構の駆動方向と平行にブロックとレールで構成されるガイド機構を設置した。また本研究ではテーブル駆動に高い再現性が求められるので高精度な位置制御、速度制御、ならびに安定したトルクを発生させられるACサーボモータを採用した。

3）仕様

完成した水平外乱刺激発生装置の大きさは、幅0.6m×長さ 1.5m×高さ 0.1m であり、最大加速度5.0m/s2、最大速度 0.5m/s、駆動範囲±0.45m（テーブルが端にあるときは 0.8m）である。装置の制御はPCで設定した加速度、速度、加速時間、変位をUSB接続でPLCに送信し、PLC→パルス発振器→サーボ機構といった流れで行う。

4）性能評価

本研究では、水平外乱刺激発生装置のテーブルが後方に駆動しているときの加速度の大きさを「加速値」、前方に駆動しているまたは減速しているときの加速度を「減速値」と定める。加速値を1.0、1.2、1.4…4.8、5.0m/s2の21段階1セットを5回計測し、無負荷時と被験者を乗せたときの加速値の違いをテーブルに加速度センサ（GYROCUBE3A　O-NAVI）計測した。加速値 4.0m/s2 のとき、無負荷時では3.47±0.10m/s2、体重60kgの被験者を乗せたときは3.71±0.03m/s2の値を示した。どの加速値でも無負荷時よりも被験者を乗せたときの方が大きい値を示した。これはサーボシステムのオートチューニング機能が、負荷が大きいときに過度に働いてしまったためだと考えられる。しかし標準偏差は0.10m/s2 以内だったため、再現性が高いことが示せた。しかし被験者ごとに加速値が変化するので、設定値の加速値と実測値に個人ごとに差があるため、本研究では設定値ではなく実測値で解析を行った。

5）水平外乱刺激時の姿勢応答実験

被験者は健常成人10名（男性7名、女性3名、21.3±1.6歳、169.2±6.9cm、体重58.8±5.4kg）で、実験について十分な説明をした後同意を得た。

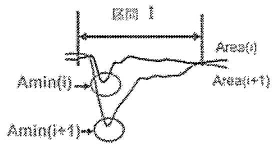
実験プロトコルは開眼・閉眼で30秒間重心動揺計（GRAVICORDER G-620　ANIMA）による計測を行った後、水平外乱刺激発生装置で加速値が[0.2m/s2 0.4m/s2 …4.2m/s2 4.4m/s2]の22段階の外乱を与えときの、足裏接地面積を足圧分布測定システム（F-SCAN NITTA）で計測した。外乱は、立位姿勢、開眼、胸の前で腕組みの状態を確認し合図をした後、1～10 秒間のうちにテーブルを駆動させ、加速値は22段階からランダムに選択した。

（倫理面への配慮）

実験の前に、実験内容を詳細に説明し、同意を得た被験者で実験を施行した。

6）解析方法

足裏接地面積は20Hzのローパスフィルタをかけた後以下の 4 つの項目を各加速値で算出した。Fig.2に接地面積の評価項目を表示した。接地面積は足裏が完全に接地しているときを 100%として解析した。面積は、利き足、非利き足、両足をそれぞれ比較した。



$$RMS(i) = \sqrt{\frac{1}{I}\int_I Area_i(t)^2}$$

$$dA\min(i) = A\min(i+1) - A\min(i)$$

$$dRMS(i) = RMS(i+1) - RMS(i)$$

Fig.2 接地面積の解析項目

解析は以下の４項目で行った

- Amin:接地面積の最小値
- dAmin:加速値を 1 段階大きくしたときの Amin の増加量
- RMS:0～1500ms間の接地面積のＲＭＳ値
- dRMS:加速値を 1 段階大きくしたときの RMS の

増加量

実際にステッピング方略を発生させた加速値で、最も小さかった値を求める。その加速値よりも 1 段階小さい加速値まででで、4 つの項目の平均値、標準偏差を求める。

C．研究結果

1）測定結果

Fig.3 に加速値 3.0m/s2 と 4.2m/s2 のときの足裏

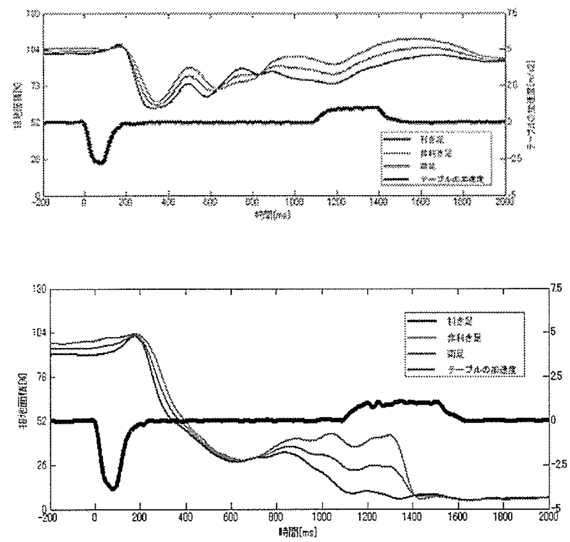


Fig.3 被験者２の足裏接地面積とテーブル加速度波形(上 3.0m/s2, 下 4.2m/s2)目

接地面積波形の典型例を示す。

Fig.3 上は、被験者は小さい加速で支持基底面内にとどまっている。まず、支持基底面積の変化は加速後に生じ、面積最低値を示し、徐々に増加していき、減速とともに増加し、静止状態の面積に戻る。利き足、非利き足の違いはほとんどない。一方、下はステッピング方略を用いた場合である。ステッピング方略を用いた加速値までは、利き足、非利き足はほぼ同期して変化している。同じ加速値を設定しても接地面積の減少のピーク値、揺れは被験者で異なった。ステッピング直前までは非利き足とほぼ同様に減少する傾向を示した。この傾向は、ステッピング方略を用いた加速値の大きさによらず全被験者で確認することが出来た。

2）評価

最も小さい加速値でステッピング方略を用いた被験者の設定値は 2.0m/s2 であった。そこで設定加速値 1.8m/s2 までの足裏接地面積データを用いて、評価指標の検討を行った。すなわち、最小加速値から 9 回の試行の平均値を算出した。その結果を Amin の平均値 (Fig.4) と、RMS値の標準偏差(Fig.5)に示す。この結果から、8 人の被験者にステッピング方略の際の加速値と一次線形的な関係が見られた。しかし、Amin の平均値では２名の被験者、RMS値の標準偏差でも別の２名の被験者がこの関係から外れた。

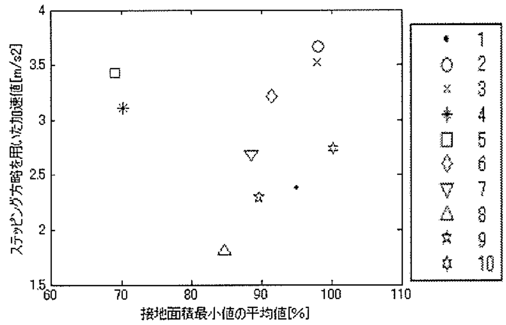


Fig.4　Amin の平均値

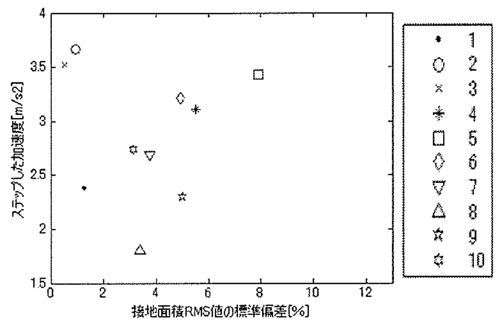


Fig.5 RMS 値の標準偏差波形(上 3.0m/s2, 下

D．考察

評価結果より、小さい加速での推定の可能性が示唆された。

Amin は接地面積の瞬間的なゆれ、RMS は区間全体のゆれを評価していると考えられる。結果から、動的バランス評価には、突発的な外乱には瞬間的に体を大きく揺らして保持しようとする要素と、小さい

外乱に対して一様に揺れているのではなく、外乱のたびに不規則な大きさで揺れることで対応しゆっくりと安定性を保持する要素の 2 つの側面があるのではないかと考えられた。また静的バランス指標（重心動揺面積、軌跡長）とステッピング方略を発生させる加速値には関係がみられなかったので、静的バランスと本研究での評価している動的バランスには関係が少ないと考えられる。

また評価に用いる設定加速値の上限を減らしていったところ、どちらも加速値 1.6 m/s2 までは一次線形的な傾向がみられたが、1.4 m/s2 以下ではその傾向はみられなかった。これらより、弱すぎる外乱では評価できない可能性と、評価に必要なサンプル数が少なかった可能性が示唆される。その一方で、大きさが異なる外乱を突発的に与えたため値が分散した可能性も考えられ、外乱の与え方も検討する必要があると考えられる。

E．結論

本研究では、弱い後方向への水平外乱を与えたときの接地面積の変化から、外乱負荷時の動的バランス評価指標を検討した。本研究結果から、弱い外乱での接地面積の最小値の平均値、ＲＭＳ値の標準偏差が、評価指標になりえることを示せた。

＜参考文献＞

なし

[1] Nashner、L.M:“Adapting reflexes controling the human posture”、 Exp Brain Res、Vol.26、pp.59-72 (1976)

[2] Nashner、L.M：“Fixed patterns of rapid postural responses among leg muscles during stance”、Exp Brain Res、Vol.30、pp.13-24 (1977)

[3] Horak、F.B.etal：'' Central programming of postural of postural movements:Adaptation to altered support-surface configurations ’、J.Neurophysiol. 55、1369-1381 (1986)

F．研究発表

１．論文発表

- なし

２．学会発表

- なし

H．知的所有権の取得状況

１．特許取得

- なし

２．実用新案登録

- なし

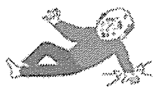
３．その他

- なし

資料：水平外乱刺激時姿勢応答からの動的バランス評価の検討

研究背景

高齢者：バランス機能の低下



転倒：滑りや躓きといった外乱が原因
→寝たきりなどの重い症状

外乱刺激に対応したバランス評価

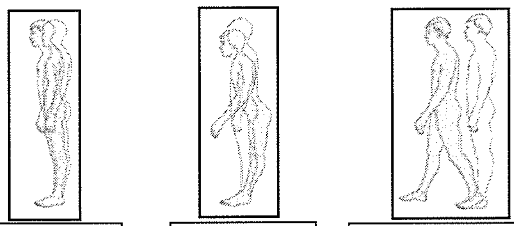
・転倒予防対策
・リハビリテーション訓練

⇩

QOL（Quality Of Life）の向上

外乱刺激時の姿勢応答

外乱の強さ



足関節方略　股関節方略　ステッピング方略

転倒を防ぐ最後の姿勢方略

・研究報告が少ない・定量的な検査方法なし
・高齢者のステッピング方略は危険が伴う

研究目的

動的バランスの基準

ステッピング方略を発生させる外乱の強さ

外乱の強さと相関性のある指標を弱い外乱での反応から検討

計測項目の選定

支持基底（面積）

定義：2足で起立したときには、
両足底面と間の部分を合計した面積

安定性

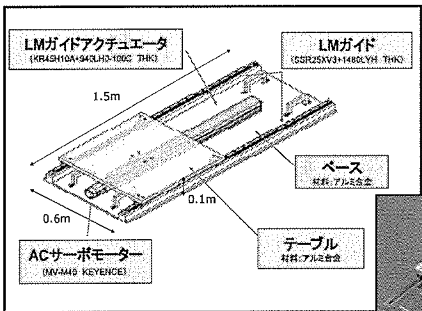
足関節方略・股関節方略→ステッピング方略

支持基底面積に変化

⇩

計測項目：両足裏接地面積

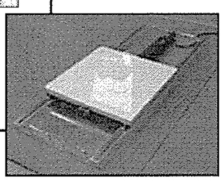
設計・開発・仕様



テーブル駆動装置の外形図

・重量：約100kg
・最大加速度：5m/s²
・最大速度：0.5m/s
・テーブル駆動範囲
：±0.45m
（端では0.8cm）

LMガイドを用いて、滑らかな動作を実現



完成した外乱刺激発生装置

性能評価

無負荷時とヒト（60kg）立位時

設定加速値[1.0m/s² 1.2m/s² 1.4m/s² ...]と
加速度センサ（GYROCUBE3A O-NAVI）の値を5回計測して比較

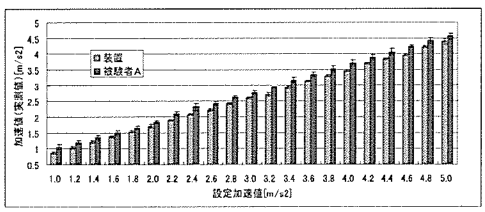


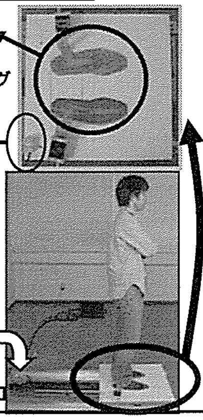
図　設定加速値と無負荷・ヒト立位時の加速値

高い再現性
解析は実測値

資料:水平外乱刺激時姿勢応答からの動的バランス評価の検討

測定システム・被験者

タクタイルセンサ
(F-SCAN NITTA)
サンプリング
100Hz
加速度センサ
(O-NAVI ONI-23503)
A/D変換
サンプリング
1000Hz
サーボ
システム



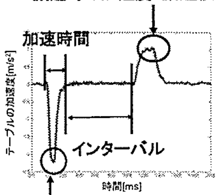
被験者

- 開眼
- 靴下着用
- 腕組み
- 男7名女3名計10名
(年齢21.3±1.6歳
身長169.2±6.9cm
体重58.8±5.4kg)
- 十分な説明の上で被験者は実験に同意

実験プロトコル

初期姿勢
合図
合図後1〜10秒後に後方外乱刺激

加速値はランダム
22回繰り返す

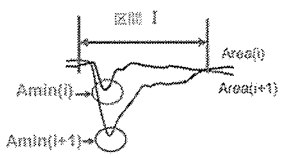


加速値:0.2〜4.4m/s²の22段階
加速時間:100ms
インターバル:1000ms
減速値:1.0m/s²

図 後方水平外乱刺激のパラメータ

解析方法(1)

$Area(i) = AREA(i)/AREA(0) \times 100[\%]$
AREA(i):足裏接地面積[cm²]
AREA(0):静止立位時の足裏接地面積の平均値[m]



$$RMS(i) = \sqrt{\frac{1}{I}\int_I Area_i(t)^2}$$

$$dA\min(i) = A\min(i+1) - A\min(i)$$

$$dRMS(i) = RMS(i+1) - RMS(i)$$

Area(i):足裏接地面積

Amin
:両足裏接地面積の最小値

dAmin
:Aminの増加量

RMS
:区間 [0〜1500ms]

dRMS
:RMSの増加量

解析方法(2)

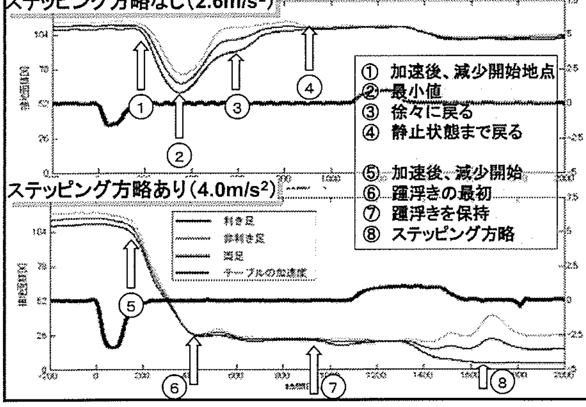
評価範囲(1)
全被験者でステッピング方略を発生させた**設定加速値の最小値**
→これよりも<u>**1段階小さい加速値**</u>

Ex)
最も小さい設定加速値でステッピング方略
被験者A 1.6m/s²
→範囲の上限:1.4m/s²
解析範囲:[0.2m/s² 0.4m/s² 0.6m/s² … 1.2m/s² 1.4m/s²]

評価範囲(2)
(1)の加速値上限を小さくして検討

測定結果〜被験者6〜

ステッピング方略なし(2.6m/s²)
ステッピング方略あり(4.0m/s²)



解析範囲

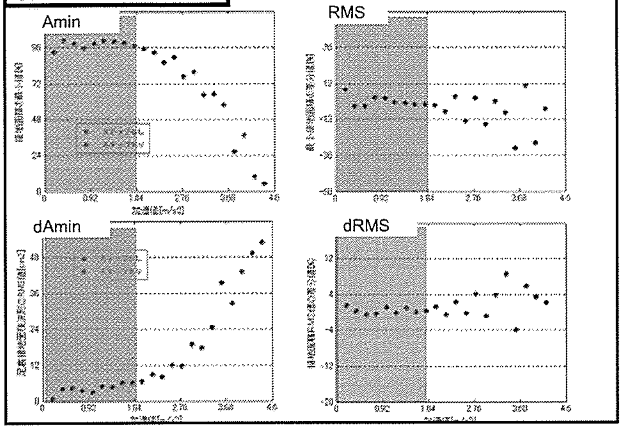
被験者	ステップした設定加速値 [m/s²]
1	2.6
2	4.0
3	3.8
4	3.4
5	3.8
6	3.6
7	3.2
8	2.0
9	2.6
10	3.0

- 最小加速値: 2.0m/s²
- 最大加速値: 4.0m/s²

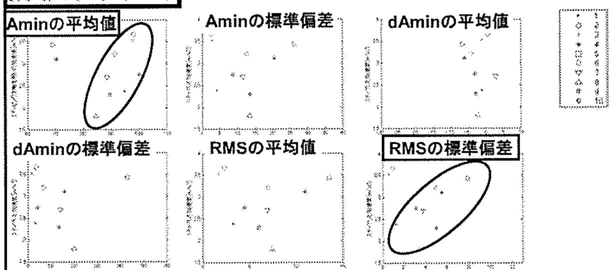
<u>解析範囲の上限</u>
1.8 m/s²

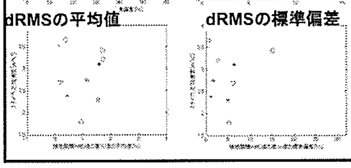
資料：水平外乱刺激時姿勢応答からの動的バランス評価の検討

解析～被験者2～



評価範囲(1)



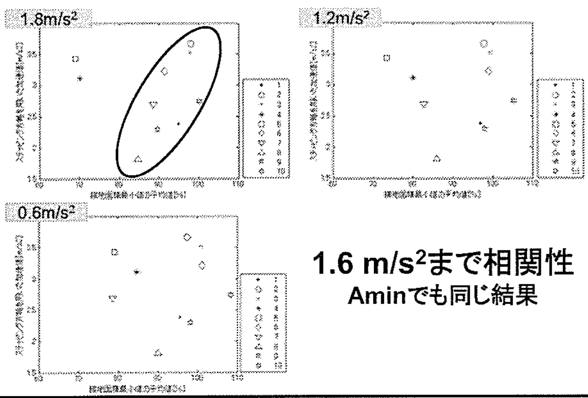


Aminの平均値
一次線形的：10人中8人

RMSの標準偏差
一次線形的：10人中8人

評価範囲(2)

Aminの平均値



1.6 m/s²まで相関性
Aminでも同じ結果

考察(1)～評価指標～

接地面積最小値(Amin)の平均値

評価指標の要素：瞬間的に体が大きく揺れること

接地面積RMS値(RMS)の標準偏差

評価指標の要素：弱い外乱でも不規則な大きさでゆっくり揺れること

考察(2)～評価に必要な外乱～

弱い外乱の検討

加速値1.4m/s²以下 ⇨ 評価×

小さすぎる外乱
少ないサンプル ⇨ 評価×

実験方法の検討

外乱 ・異なる大きさ
・突発的 ⇨ 値の分散

外乱負荷方法の検討の必要性

まとめ

動的バランス評価指標
ステッピング方略した外乱加速値と相関性のある指標

弱い後方向の水平外乱刺激時の
足裏接地面積の変化から検討

⇩

評価指標

加速値1.6m/s²までの最小値の平均値
加速値1.6m/s²までのRMS値標準偏差

検討課題
・1.4m/s2以下での評価・必要なサンプル数
・外乱負荷の方法

III.　研究成果の刊行に関する一覧表

研究成果の刊行に関する一覧表

雑誌

発表者氏名	論文タイトル名	発表誌名	巻号	ページ	出版年
Ichinoseki-Sekine N, Kuwae Y, Higashi Y, Fujimoto T, Sekine M, Tamura T	Improving the accuracy of pedometer used by the elderly with the FFT algorithm	Medicine & Science in Sports & Exercise	38(9)	1674-1681	2006
山本弘毅, 吉村拓巳, 関根正樹, 田村俊世	高齢者のバランス機能改善を目的とした足裏刺激装置の開発	第21回生体・生理工学シンポジウム論文集		413-414	2006
関根正樹, 木内尚子, 前田祐佳, 田村俊世, 桑江　豊, 東　祐二, 藤元登四郎, 大島秀武, 志賀利一	高齢者の歩容に対応した歩数計の開発ーカウントアルゴリズムの検討ー	第21回生体・生理工学シンポジウム論文集		521-522	2006
吉村拓巳, 山本弘毅, 関根正樹, 田村俊世	転倒エアバッグ開発のための転倒検出方法の検討	第21回生体・生理工学シンポジウム論文集		523-524	2006

IV.　研究成果の刊行物・別刷

Improving the Accuracy of Pedometer Used by the Elderly with the FFT Algorithm

NORIKO ICHINOSEKI-SEKINE[1], YUTAKA KUWAE[2], YUJI HIGASHI[2], TOSHIRO FUJIMOTO[2], MASAKI SEKINE[3], and TOSHIYO TAMURA[3]

[1]*School of Science and Engineering, Tokyo Denki University, Saitama, JAPAN;* [2]*Fujimoto Hayasuzu Hospital, Miyazaki, JAPAN; and* [3]*Department of Medical System Engineering, Faculty of Engineering, Chiba University, Chiba, JAPAN*

ABSTRACT

ICHINOSEKI-SEKINE, N., Y. KUWAE, Y. HIGASHI, T. FUJIMOTO, M. SEKINE, and T. TAMURA. Improving the Accuracy of Pedometer Used by the Elderly with the FFT Algorithm. *Med. Sci. Sports Exerc.*, Vol. 38, No. 9, pp. 1674–1681, 2006. **Purpose:** The aim of this study was to investigate and improve the accuracy of accelerometer-type pedometers used by the elderly with slow walking speeds, with or without gait disorders, who do or do not use a cane. **Methods:** Eighteen subjects walked with a cane (5 males, 13 females; age, 80.9 ± 7.7 yr; height, 148.1 ± 7.7 cm; weight, 51.8 ± 8.8 kg (mean ± SD); nine had impaired gait), and 31 subjects walked without a cane (7 males, 24 females; age, 80.9 ± 7.7 yr; height, 148.1 ± 7.7 cm; weight, 51.8 ± 8.8 kg; 15 had impaired gait). Subjects walked for approximately 20 m (10 m in each direction and a turning arc) at their own speed. We determined the number of steps by pedometer (PM), by visually counting the actual number of steps (RM), and by the triaxial acceleration signals. The power spectrum of the accelerometer in each direction calculated by fast Fourier transform (FFT) for a 4-s temporal window was normalized with the maximum power of each window. It was composited, and the frequency at maximum power was considered as the cadence. The number of steps taken (FM) was determined by summing all the estimated steps in each window. **Results:** PM was significantly less than the RM ($P < 0.05$), and the error of PM was 53.2 ± 34.1% of RM. FM did not differ from the RM, and the average error of FM was −0.7 ± 7.9% of RM (absolute value: 5.8 ± 5.3%). **Conclusion:** We suggest that our FFT method is suitable for estimating the number of steps during walking in this population. **Key Words:** ACCELERATION, CADENCE, FAST FOURIER TRANSFORM (FFT), NUMBER OF WALK STEPS

Promoting increased physical activity among elderly patients is needed to improve their quality of life (QOL). A physically active lifestyle is known to reduce the risk of various chronic diseases such as coronary artery disease, diabetes mellitus, hypertension, and obesity (2). Likewise, regaining the ability to walk and improving the performance of activities of daily living (ADL) are major goals of rehabilitation in patients with walking disorders (28). However, nearly two thirds of stroke survivors have impaired mobility and assume a sedentary lifestyle (29), which increases the risk for recurrent cardiovascular events and stroke (20–22). Thus, the measurement of physical activity, specifically locomotion, in the elderly and disabled populations would help identify individuals at risk and would also encourage walking.

A few methods for measuring physical activity include observational measures, self-reporting or diaries, and pedometers. Self-reporting has been used historically but is limited by the difficulty of estimating total daily walking distance (1), variation with scoring procedures (23), and dependence on patients' memory. Because a pedometer is easy to use and relatively inexpensive, this device is widely used to determine daily activity levels (5,27). The pedometer is a simple device that directly measures physical activity (e.g., number of walking steps, distance, time, etc.) and provides immediate feedback to participants and researchers. Pedometers commonly use two methods for counting steps; one is based on the movement of a mechanical pendulum, and the other method uses a piezoresistive accelerometer and a threshold of acceleration signals (17). A mechanical pedometer is inexpensive (the lowest price in Japan is approximately \$1) and is easy to use. Daily walking steps and exercise habits are known to be related to peak oxygen consumption and ventilation (30). The acceleration pedometer can also be inexpensive (approximately \$30–\$300; the standard selling price in Japan is approximately \$50), is easy to use, and is one of the most commonly used types of pedometer. The

Address for correspondence: Noriko Ichinoseki-Sekine, Ph.D., Institute of Health and Sports Sciences, Juntendo University, 1-1 Hiragagakuendai, Inba, Chiba 2701695 Japan; E-mail: n.sekine@ieee.org.
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TABLE 1. Physical characteristics of subjects (mean ± SD).

	SC	IC	SN	IN
	N = 9 (male 2, female 7)	N = 9 (male 3, female 6)	N = 16 (male 3, female 13)	N = 15 (male 4, female 11)
Age (yr)	80.7 ± 9.7	78.1 ± 6.4	81.5 ± 9.5	82.0 ± 4.8
Height (cm)	149.2 ± 10.1	149.2 ± 5.9	147.2 ± 6.4	147.8 ± 8.8
Weight (kg)	48.2 ± 9.0	53.0 ± 7.1	51.7 ± 8.8	53.3 ± 9.8
BMI ($kg \cdot m^{-2}$)	21.5 ± 2.8	23.7 ± 1.8	23.8 ± 3.4	24.3 ± 3.4

SC, senile gait with cane; IC, impaired gait with cane; SN, senile gait without cane; IN, impaired gait without cane.

accelerometer output has a good relationship to energy consumption (6,13,19), which is widely accepted as the standard reference for physical activity.

We examined pedometer accuracy in this study because of the increasing interest in using pedometers as a means of promoting increased physical activity (26,27). Several studies have demonstrated substantial variations in the accuracy of pedometers among brands and methods (4,7,14,18,24). In each of these studies, all the pedometers undercounted the number of steps by approximately 50–90% at walking speeds of 50–54 $m \cdot min^{-1}$ (approximately 3.0–3.2 $km \cdot h^{-1}$). However, most of the pedometers accurately counted the actual steps at walking speeds faster than 94 $m \cdot min^{-1}$ (approximately 5.6 $km \cdot h^{-1}$). At a slow walking speed of 1.0 mph (approximately 27 $m \cdot min^{-1}$, or 1.6 $km \cdot h^{-1}$), the acceleration pedometer (with a 56% accuracy rate) was superior to the mechanical pedometer, which indicated 7–20% of the actual number of steps. These studies suggested that pedometers may undercount the number of walking steps in individuals who walk slower than 2 mph (approximately 53 $m \cdot min^{-1}$, or 3.2 $km \cdot h^{-1}$).

As indicated above, the accuracy of pedometers that depend on vertical movement is less for subjects who walk slowly. Besides, elderly patients and patients who require rehabilitation training to recover their gait walk slowly (16). In addition, walking aids are often used during gait training to increase stability, reduce the risk of falling, and improve independent walking (2,12). Rehabilitation for improving ADL requires accurate assessment; thus, a pedometer that accurately counts the number of walking steps is a useful device. However, to our knowledge, pedometers are inaccurate when used with patients who have a slow walking speed, have an impaired gait, or use a walking aid such as a cane (9). An ankle-worn accelerometer-based pedometer was found to be accurate in patients with gait disorders caused by conditions such as stroke (16), but this device is expensive (approximately $1500) and difficult to use compared with conventional pedometers because it requires a computer to operate.

The aim of this study was to assess and improve the accuracy of pedometer counts for people who are receiving gait training and have a slow walking speed, heterogeneity in their gait pattern, or ground reaction force (reaction force upward when foot is on the ground) asymmetry because of aging, gait disorders, or the use of a cane. A step frequency (cadence) obtained from acceleration signals during walking was used to estimate the number of walking steps because vertical movement was considered to be small in this population. The number of steps counted by the pedometer method (PM) and the number of steps estimated by using fast Fourier transform (FFT; FFT method, FM) were compared with

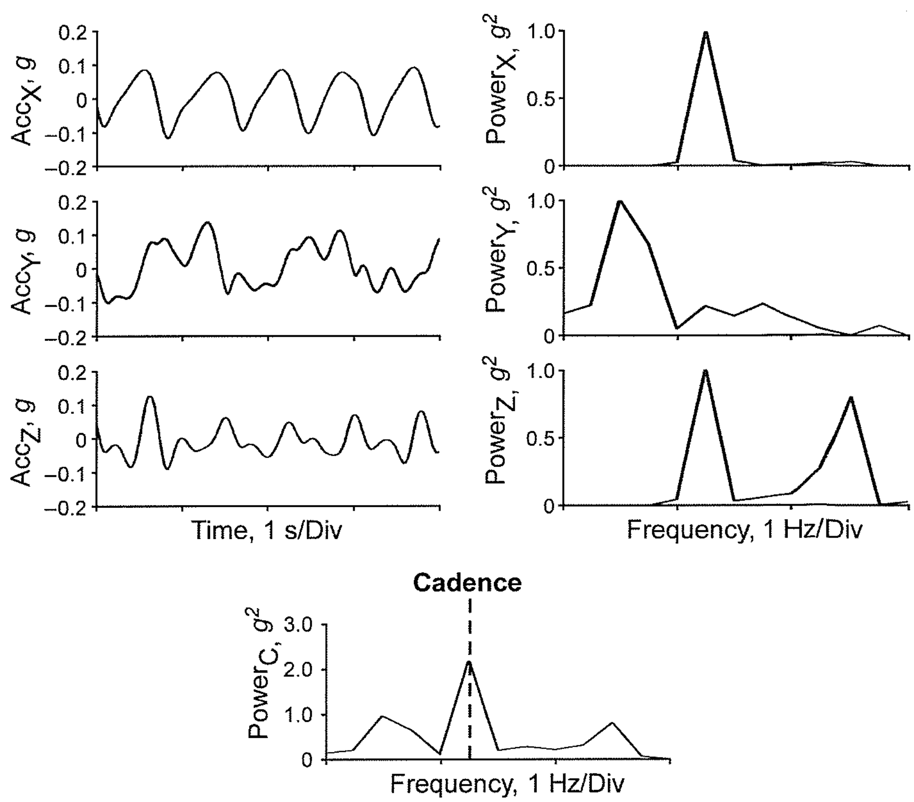

FIGURE 1—Typical examples of the acceleration signals, Acc_X, Acc_Y, Acc_Z, and their normalized power spectrum, $power_X$, $power_Y$, and $power_Z$. The frequency at the maximum power of the composite power spectrum, $power_C$, was considered as the cadence of each window.

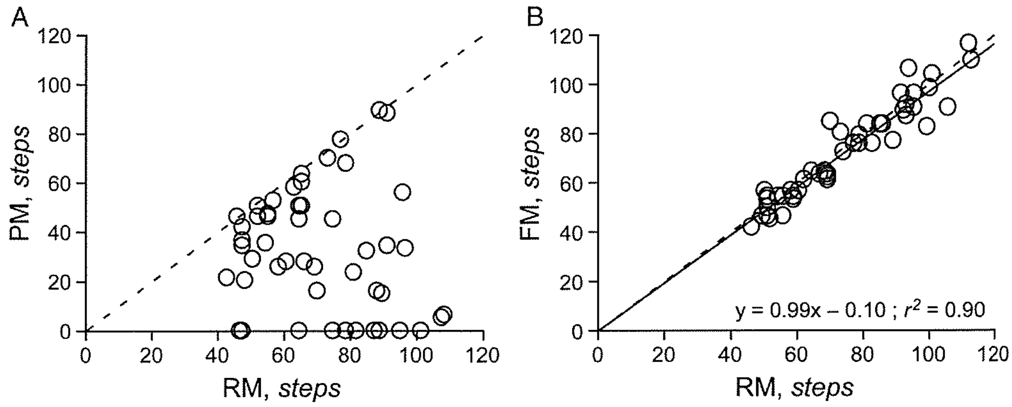

FIGURE 2—Relationship between number of steps counted visually by a physical therapist (RM) and A) number of steps counted by pedometer (PM) and B) number of steps estimated using the FFT algorithm (FM). FM significantly correlated with the actual number of steps ($P < 0.0001$).

the actual number of steps visually counted by a physical therapist (reference method, RM).

METHODS

Subjects. Forty-nine subjects (12 males, 37 females; age, 80.9 ± 7.7 yr; height, 148.1 ± 7.7 cm; weight, 51.8 ± 8.8 kg; BMI, 23.5 ± 3.1; mean ± SD) from the commuting rehabilitation services center at Fujimoto Hayasuzu Hospital, Japan, participated in this study (Table 1). Eighteen subjects walked with a cane (senile gait (SC), N = 9; impaired gait (IC), N = 9), and 31 subjects walked without a cane (senile gait (SN), N = 16; impaired gait (IN), N = 15). Senile gait was defined as the gait disturbance that often occurs in the elderly, not originated from diseases. The various gait-impaired subjects suffered from hemiplegia, Parkinson's disease, degenerative joint disease, lumbar spinal canal stenosis, and total knee arthroplasty. There were no major physical differences among the four groups. This study was approved by the ethics committee of Fujimoto Hayasuzu Hospital, and written informed consent was obtained from all subjects.

Measurement system. An acceleration pedometer and an acceleration measurement system were used in this study. The pedometer was a biaxial acceleration pedometer (HJ-720IT, Omron Healthcare Co., Ltd., Kyoto, Japan; size, 73 × 47 × 16 mm; weight, 37 g) based on a threshold of acceleration signals. The acceleration measurement system consisted of an accelerometer device and a telemeter system (WEB-5000, Nihon Kohden Co., Ltd., Tokyo, Japan). The accelerometer device consisted of a triaxial piezo accelero meter (Akebono Brake Industry Co., Ltd., Tokyo, Japan), amplifiers, and low-pass filters (size, 30 × 40 × 20 mm; weight, 20 g; range, ± 2 g; frequency response, 0–100 Hz, cutoff frequency, 50 Hz; 1 $g \approx 9.8$ m·s^{-2}). We recorded raw acceleration signals in the anteroposterior (x), lateral (y), and vertical (z) directions. The accelerometer outputs were digitized and recorded at a sampling rate of 128 Hz.

Experimental design. We recorded height and weight measurements for the subjects before testing. Subjects walked for approximately 20 m (10 m in each direction and a turning arc) at their own speed. After the accelerometer device was calibrated by measuring the outputs under a controlled inclination (10), the device was fixed on an acrylic plate that had two slits for a waist belt. Using an elastic waist belt, the accelerometer device was attached to the back of the lumbosacral region of the vertebral column of the subject, which was close to the subject's center of gravity. The pedometer was also attached to the right dorsal region by the same waist belt. We determined the number of steps counted by the pedometer (PM), the actual number of steps visually counted by a physical therapist using a hand-tally counter (RM), and the raw triaxial acceleration signals for the entire walking period. The walking time for the 10-m leg of the route was also measured by a stopwatch.

Estimation of the number of steps using FFT. We estimated the number of steps using the power spectrum of the raw acceleration signals for the three directions (FM, Fig. 1). The power spectrum of each direction in the range of 0.5–3.0 Hz was calculated by FFT for a temporal window that contained 512 samples (4 s each) of the signal. It was normalized with the maximum power of each window, and the normalized power spectrums of the three directions were composited. We considered the frequency at the maximum power in the composite power spectrum as the cadence, and the number of steps at the window were estimated by multiplying the cadence and 4-s window length. This process was repeated in the next nonoverlapping temporal window. Finally, the number of steps of the whole walking period was obtained by summing all the estimated number of steps in each window. This estimation procedure was performed on MATLAB, Version 6.0 (MathWorks, Inc., Natick, MA).

TABLE 2. RM, PM, FM, and walking speed (mean ± SD).

	SC (N = 9)	IC (N = 9)	SN (N =16)	IN (N = 15)
Number of steps (steps)				
RM	80.3 ± 14.5	78.3 ± 19.1	74.2 ± 17.5	69.0 ± 21.3
PM	33.0 ± 18.6**	26.7 ± 28.1**	36.9 ± 27.5**	30.4 ± 23.5**
FM	81.4 ± 13.5	78.6 ± 21.1	72.1 ± 18.7	68.7 ± 21.6
Walking speed (m·min^{-1})	35.0 ± 11.5	33.0 ± 12.8	44.1 ± 11.2	41.2 ± 7.2

SC, senile gait with cane; IC, impaired gait with cane; SN, senile gait without cane; IN, impaired gait without cane; RM, number of steps counted visually by a physical therapist; PM, number of steps counted by pedometer; FM, number of steps estimated using the FFT algorithm. ** $P < 0.01$ vs RM.

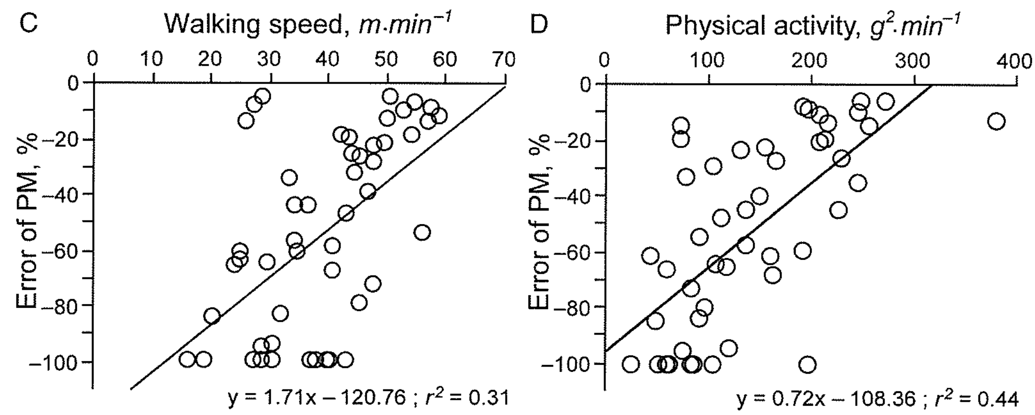

FIGURE 3—Relationship between the error of PM (% RM) and C) walking speed and D) physical activity in the vertical direction. The error of PM was significantly correlated with the walking speed and physical activity (P < 0.0001).

Data analysis and statistics. The errors of PM or FM (%) were defined as 100 × (PM or FM − RM)/RM. To assess the relationship between the physical activity and counting step ability, the power of the raw acceleration signal (P) associated with the physical activity (19) was estimated by the following equation:

$$P = \|x(n) - \bar{x}\|_2^2$$

where x(n) is the raw acceleration signal and $\bar{x}$ is the mean of x(n). We defined the power of entire walking period per minute as the physical activity.

We compared the error of FM to the error of PM. One-way analysis of variance (ANOVA) was used to analyze differences among the four groups classified by walking condition, and differences between two errors were evaluated by paired or nonpaired t-test. Significant F-values were followed up with Scheffe's *post hoc* test. Correlation analysis was used to quantify the association between values. A value of $P < 0.05$ was considered statistically significant, and statistical analysis was performed using StatView, Version 5.0 (SAS Institute Inc., Cary, NC).

RESULTS

The walking speed for the 10-m leg of the route and RM during the 20-m out-and-back course in this study were 39.5 ± 11.1 m·min^{-1} (approximately 2.4 km·h^{-1}; range: 16.6–59.5 m·min^{-1}, approximately 1.3–3.6 km·h^{-1}) and 74.5 ± 18.5 steps (range: 46–113 steps), respectively. The number of walking steps of all subjects was undercounted by the pedometer and did not correlate significantly (Fig. 2A). There was no significant difference between the RM and FM for all the subjects. Under the walking condition, there was no significant difference across SC, IC, SN, and IN subjects in RM, FM, and walking speed (Table 2). However, there were significant differences between the RM and PM in each group.

The pedometer produced values that were more than 90% of RM in four subjects (well-counted subjects) and that differed by 100% in 10 subjects (uncounted subjects (SC), $N = 1$; IC, $N = 3$; SN, $N = 3$; IN, $N = 3$). The average error of PM for all the subjects was −53.2 ± 34.1% of the RM (range: −5.4 to −100%) and significantly correlated with the walking speed (r = 0.56, $P < 0.01$) and triaxial physical activities (anteroposterior: r = 0.66; lateral: r = 0.58; vertical: r = 0.68, $P < 0.01$; Fig. 3). The average of the error of PM in the four groups was more than −40% (Fig. 4). There was no significant difference among the errors of PM in the four groups, SC, IC, SN, and SN. The error of PM showed a significant correlation (r = 0.58, $P < 0.05$) with the walking speed in the SN group but not in the other three groups. In contrast, the errors of PM in the SC, IC, and SN groups were significantly correlated with the physical activity in the vertical direction (SC: r = 0.83; IC: r = 0.79; SN: r = 0.71; $P < 0.01$). When all the subjects were divided into two groups based on an average walking speed of 40 m·min^{-1} (approximately 2.4 km·h^{-1}), significant

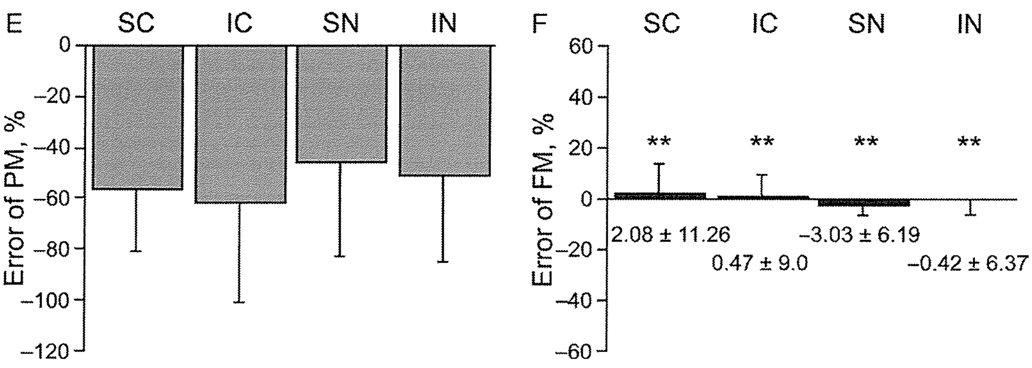

FIGURE 4—The errors of PM (% RM) (E) and the error of FM (F) for senile gait with cane (SC), impaired gait with cane (IC), senile gait without cane (SN), and impaired gait without cane (IN). ** P < 0.01 vs the errors of PM within the same group.

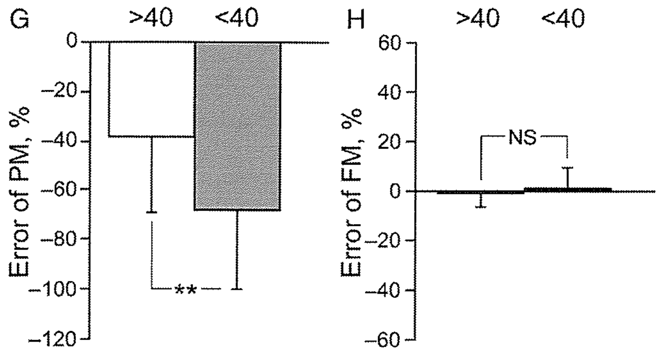

FIGURE 5—Effect of speed on the error of PM (% RM) (G) and the error of FM (H). > 40, greater than 40 m·min^{-1} average walking speed group, N = 26; < 40, less than 40 m·min^{-1} average walking speed group, N = 23. ** P < 0.01; NS: not significant.

differences in the errors of PM between the fast (> 40 m·min^{-1}, N = 26) and slow (< 40 m·min^{-1}, N = 23) groups were observed (Fig. 5).

The average physical activities for all the subjects were 76.7 ± 31.4 g^2·min^{-1} in x, 87.3 ± 48.4 g^2·min^{-1} in y, and 139.0 ± 76.9 g^2·min^{-1} in z directions. There was no significant difference among the four groups in the average value of physical activity. The average value of FM in this study was 74.0 ± 19.3 steps and did not differ from the RM, but did significantly differ from PM (P < 0.01). Additionally, there was no significant difference between the FM and RM in any group (Table 2). The FM was significantly correlated with the RM (r = 0.95, P < 0.01, Fig. 2B). The average error of FM was −0.7 ± 7.9% of RM (the average absolute value of the error of FM was 5.8 ± 5.3%); this was significantly negative when correlated with the physical activity in the anteroposterior (r = −0.30, P < 0.05) and vertical (r = −0.35, P < 0.05) directions, but was not correlated with the walking speed (Fig. 6). The errors of FM for the four groups were significantly different from the errors of PM (P < 0.01; Fig. 4). The errors of FM between the fast and slow groups did not differ significantly (Fig. 5).

DISCUSSION

Le Masurier and Tudor-Locke et al. (15) examined two different types of pedometers at five different treadmill speeds and found that the magnitude of the error might not be an important hindrance to the assessment of healthy adults. However, they noted that the magnitude of the error would be a problem when monitoring elderly people with slow walking speeds. In a recent study, several pedometers were tested in self-selected speed experiments in healthy adults, and the average walking speed was approximately 96 m·min^{-1} (24). Most of the pedometers showed poor accuracies under low walking speeds in previous studies (7,15). In this study, the walking speed was extremely slow compared with previous studies; the average walking speed was 39.5 m·min^{-1} and ranged from 16.6 to 59.5 m·min^{-1}. Our results supported the results of previous studies because the pedometer underestimated the number of steps; the error of PM was highly correlated with the walking speed, which means that the error of PM increases with decreases in the walking speed. Several reports only examined the pedometers on straight walking courses or treadmills (8); however, in this study, our walking course included out-and-back segments and a turning arc. Thus, there was a time period before a steady walking rate was reached. Moreover, our subjects included patients with gait disorders and/or those who required the use of a cane during walking. Therefore, the experimental design in this study is considered to represent a very severe situation for the evaluation of pedometer.

Under the conditions of this study, the acceleration-type pedometer could not count the number of steps accurately. There were 10 uncounted subjects who were distributed over the four groups. Significant difference was not observed among the four groups with regard to the error of PM; thus, using a cane and having an impaired gait did not effect the accuracy of the pedometer in this study. Figure 7 shows typical examples of the raw acceleration signals in the three directions. The black line represents a

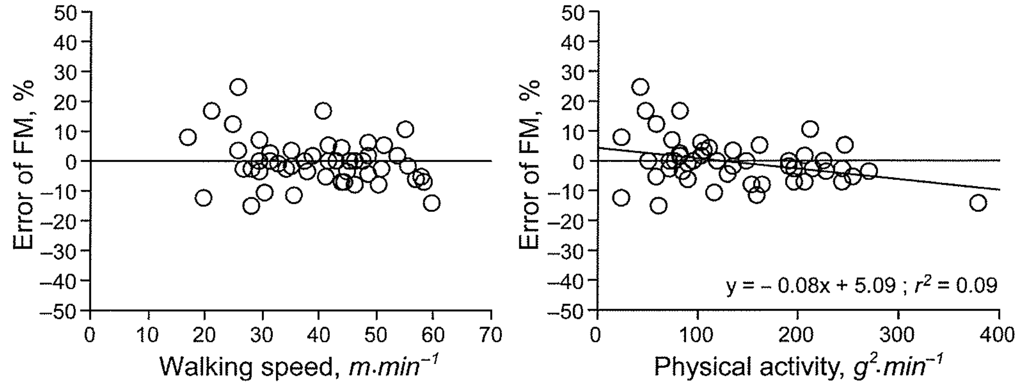

FIGURE 6—Relationship between the error of FM and (I) walking speed and (J) physical activity in the vertical direction. The error of FM was significantly correlated with the physical activity (P < 0.05).

well-counted subject, and the gray line represents an uncounted subject. When the amplitude of the acceleration was small, as with the gray line, the pedometer could not count the steps, which suggests that a counting method that relies on a threshold of acceleration in the vertical direction is not suitable for this population. The relationship between the counting error and the physical activity in the vertical direction also suggests the difficulty in counting based on movement in the vertical direction.

To improve the accuracy of the pedometer, we attempted to estimate the number of steps using the cadence obtained from the acceleration signals in the three directions. The estimation results of our FFT method were accurate in the elderly who walked slowly, with or without gait disorders, and with or without using a cane: the average error of FM was $-0.7 \pm 7.9\%$ of the RM. By using FFT, we were able to obtain the cadence from the small amplitude of the acceleration as shown in Figure 7; thus, our FFT method was able to estimate the number of steps in the subjects whose steps were not counted by the pedometer. Moreover, our FFT method estimated the number of steps regardless of the presence of a gait disorder. Generally, these populations are not conducive to a counting method based on the threshold of vertical movement. Furthermore, our method is amenable to the attachment of the device. A previous study reported that the accuracy of pedometers, especially mechanical ones, is affected by pedometer tilt (7). Our FFT method will not limit the placement of the device or device position on the waist belt because it is based on composite acceleration signals in the three directions. Thus, our FFT method might be effective in estimating, with high accuracy, the number of steps in overweight and obese adults or elderly patients with kyphosis.

Nevertheless, the number of steps was under-and over-estimated by our FFT method. Only one subject's error of FM was poorer than that of PM. The number of steps of this subject was underestimated by 14.3% of RM by our FFT method, whereas it was undercounted by 12.5% of RM by the pedometer. The error of FM in this subject occurred during the turning arc; the cadences of the 4-s windows in the three directions were not in synchrony, and the physical activities in the three directions were large compared with other subjects. However, nine subjects showed errors of FM that were larger than 10% of RM. The amplitudes of the errors of FM were smaller than those of the pedometers individually except in one subject, whose error of FM exceeded 20% of the RM; however, it was better than the results of the pedometer. The subject's number of steps was undercounted by 60% of RM by the pedometer. Therefore, we suggest that our FFT method will be able to provide the number of steps accurately even in elderly subjects with slow walking velocities, gait disorders, and cane use during walking.

A 4-s window length was used for calculation of cadence in this study because none of the subjects took more than 4-s for a single step. Using a 4-s window length meant that there was a 4-s delay at each measurement period. In the present study, the time period at the end of the measurement, which was not a complete 4-s length, was cut off. Consequently, there is a possibility that this cutoff time period caused some estimation error. The 4-s window length was considered suitable for our subjects, but the window length must be chosen carefully based on the characteristics of the subjects. Another point that should be considered is the handling of the turning arc. The amplitudes of the acceleration signals during the turning arc differed from those of walking segment; thus, there was a tendency for the determined cadence at each window to be low when the 4-s window included both the walking and turning periods. However, a high-frequency acceleration signal that did not originate from the walking movement sometimes appeared at the turning period. These problems must be resolved for a more accurate estimation.

In this study, we attempted to estimate the number of steps during the walking and turning segments. As mentioned above, the results of this study showed that our FFT method is suitable for people who must undergo gait training. Maintaining patient motivation is important for reaching rehabilitation goals and regaining QOL. Our method will assist with efforts to maintain patient motivation

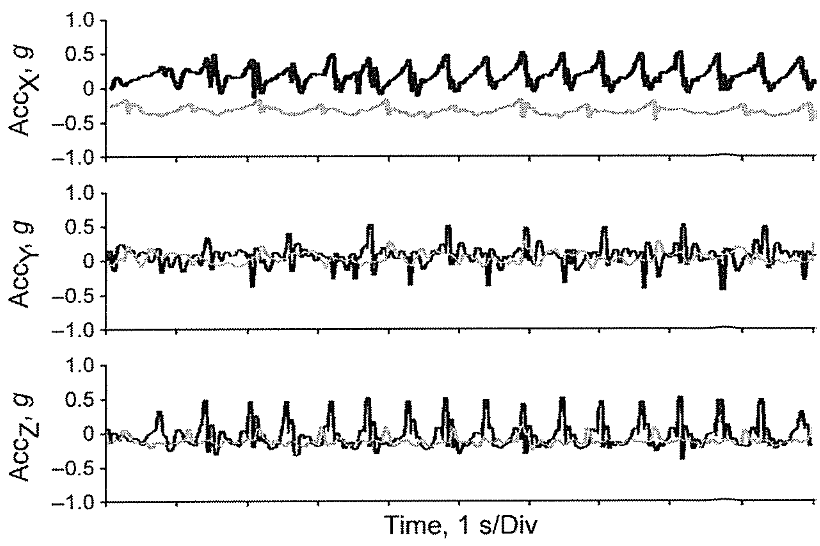

FIGURE 7—Typical examples of the acceleration signals Acc_X, Acc_Y, and Acc_Z in well-counted subjects (PM was more than 90% of RM; *black line*) and uncounted subjects (PM was 0% of RM; *gray line*).

because the use of the pedometer will increase the motivation of the patient (4). Consequently, our FFT method is useful for promoting the continuation of gait training, which can help patients improve their muscle strength, avoid falls, and regain the ability to perform ADL. In addition, our method will reduce the workload of the therapist. For the application to clinical situations or daily life, price and ease of use are important factors. A digital signal processor (DSP) or simple low-power CPU is needed to calculate a FFT; however, a pedometer with a DSP could be made that would not be prohibitively expensive (approximately $200) and that would be as easy to use as current pedometers. In the future, improved device and time-frequency analysis may be available for application under free-living conditions (3,11,25); this is required for increasing and promoting daily physical activity, although we must point out the changes in speed inside the house.

In summary, we assessed the accuracy of a pedometer in elderly subjects with regard to walking speed, the presence of a gait disorder, and the use of a cane. We also estimated the number of steps using the cadence obtained from the composite power spectrums of triaxial acceleration signals using a FFT algorithm and compared this with the number of steps visually counted by a therapist. The pedometer was inaccurate in our subject population, whereas our FFT method was reasonably accurate at estimating the number of steps. The number of steps estimated by our FFT method was significantly correlated with the actual number of steps, and the average value of the estimation error was $-0.7 \pm 7.9\%$. Our FFT method needs improvement because estimation errors of more than 10% existed. We suggest that our method is adequate and suitable for estimating the number of steps during walking in elderly people with slow walking speeds, with or without gait disorders, and with or without the use of a cane.

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高齢者のバランス機能改善を目的とした足底刺激装置の開発
Development of an insole stimulation device for improving the balance function in the elderly

山本 弘毅[1], 吉村 拓巳[2], 関根 正樹[3], 田村 俊世[3]
Hiroki YAMAMOTO[1], Takumi YOSIMURA[2], Masaki SEKINE[1], Toshiyo TAMURA[1]
[1] 千葉大学大学院 自然科学研究科, [2] 東京都立産業技術高等専門学校, [3] 千葉大学 工学部
[1] Graduate School of Science and Technology, Chiba University,
[2]Tokyo Metropolitan College of Technology, [3]Faculty of Engineering, Chiba University

Abstract Many elderly people face the risk of falling down because their balance function decline. We made vibrating insoles for the purpose of improving the balance function. The acceleration signals were measured while the healthy subjects were walking with and without stimulation. Autocorrelation, root mean square value and trajectory area were calculated from the acceleration signals. We noted that root mean square value, trajectory area and the force of heel lift were decreased during stimulation. Vibrating insoles could reduce the sway of the body, and the probability of improving balance function was suggested.

1. はじめに

近年，社会の高齢化が進んでいる。高齢者は、加齢に伴う身体機能の低下、特にバランス機能の低下により転倒の危険性が高くなる[1]。転倒により，日常生活動作（ADL：Activities of Daily Living）の狭小化，寝たきり状態への移行が進むと考えられる。転倒を防止するためにはバランス機能の維持が重要である。バランス機能の低下防止においては運動面（筋骨格系，神経筋協同収縮系など）への刺激が注目されがちであるが，本研究では，その刺激入力の容易さから，感覚面（視覚系，前庭迷路，体性感覚系など）の中でも，体制感覚への刺激入力に着目した。体制感覚の刺激によるバランス機能の改善に対して、静的な立位ではあるが足底の刺激が有効であることが報告されている[2]。本研究では，歩行時のような動的なバランス機能に対しても足底の刺激が有効であると仮定して，ウェアラブルな足底刺激装置を試作し，その効果を検証した。

1.1 振動刺激装置

開発した足底に刺激を与える装置（以下振動刺激装置）の外観を Fig.1 に示す。本研究では，足底刺激を行うための装置としてフレキシブル基板上に回路をプリントし，回路上にタクトスイッチ（SKQGAD, ALPS）を実装，その上に振動モータ（FM34F, 東京パーツ工業）を取り付けた。

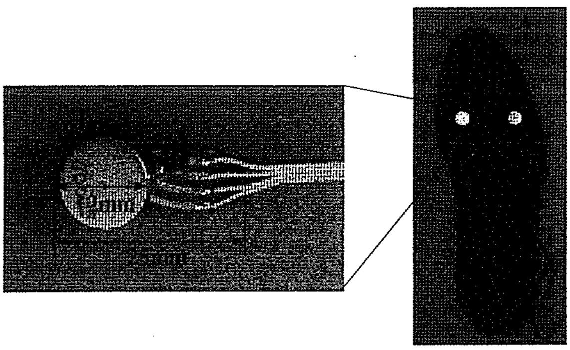

Fig.1 vibration system

この振動刺激装置を，刺激を行う箇所に穴をくり抜いたスポンジに取り付けることで，インソール型の足底刺激装置とした。振動刺激装置はスイッチが ON になることでモータが ON になる設計としており、立脚時に足底に振動刺激を与えるようになっている。今回のように，振動モータをインソールと独立して作製することで，インソールの穴の箇所を変えることで刺激を行う箇所を容易に変えることが可能である。

1.2 実験方法

1.2.1 測定装置

本研究では加速度により歩行時のバランス機能の評価を行った。加速度センサ（H48A 日立金属）からの信号はマルチテレメータシステム（WEB5000, 日本光電）によって収集され，A/D 変換ボードを介し PC に取り込まれ，サンプリング周波数 1[kHz]で記録された。

1.2.2 測定方法

加速度センサは被験者の腰背部中央に取り付けた。測定中にセンサがずれないように，センサをベルトによって固定した。テレメータの送信部は歩行の邪魔にならないようリュックの中に入れ，それを背負って測定を行った。 加速度は前後，左右，上下方向の 3 軸について測定した。

測定は刺激無しの場合と両足に刺激を与えた場合の2種類の歩行に対して行った。歩行は直線 10[m]程度を往復とした。測定中に与えられる振動刺激の箇所は，踵骨と第一中足骨、第五中足骨付近の 3 箇所であり，歩行中の立脚時に刺激が与えられた。

測定対象者は正常歩行を行える健常成人男性 6 名（年齢 24.8±3.3 歳，身長 172.7±2.5cm，体重 62±7.9kg）とした。測定中の履物は，履物の種類による加速度への影響を考慮して，全員こちらで用意したものを使用した。本研究は千葉大学工学部倫理規定に沿って行われ、各被験者には事前に実験方法を説明し承諾を得ている。

1.2.3 解析方法

測定で得られた加速度波形の解析は以下の方法で行った。刺激なしと刺激ありの両歩行で得られた波形における両足 5 歩分の波形を取り出し，自己相関，二乗平均平方根（root mean square, 以下 rms 値）を計算した。取り出したデータは歩行のはじめと終わりを除く安定した波形を切り出した。

また、rms 値の結果から前後左右方向の加速度の軌跡面積を検討した。軌跡面積は前後方向と左右方向の rms 値を掛け合わせて求めることができる。

2. 結果

計算の結果，自己相関では変化は見られなかったが，rms値で刺激無しの歩行と刺激ありの歩行に差が見られた。Fig.2 に各被験者の刺激ありの歩行時における加速度 3 軸の rms 値を示す。刺激無しの歩行時の rms 値を 1 として正規化した値を示す。

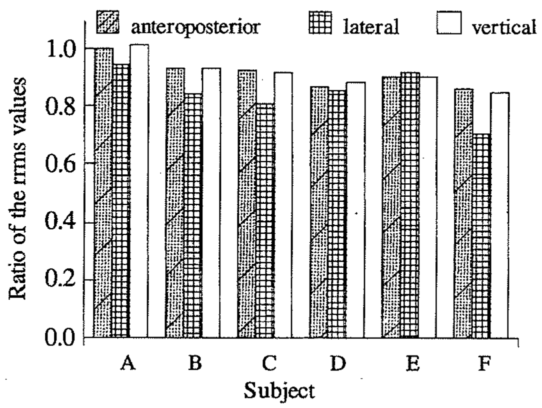

Fig.2　Ratio of root mean square values without stimulation against to with stimulation

被験者Aを除いて全体的にrms値が減少していることが確認できた。

前後左右方向の刺激なしと刺激ありの場合の加速度の軌跡の例を Fig.3 に示す。1[g]=9.8[m/s^2]とする。

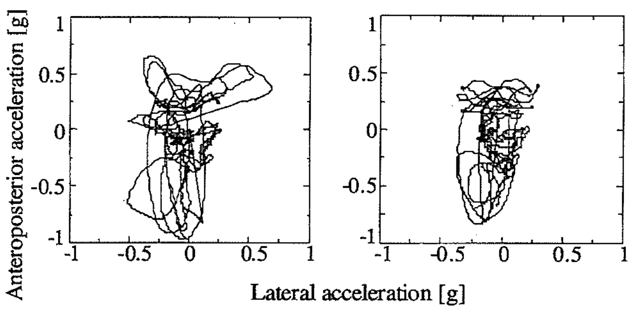

Fig3.　Acceleration trajectory
(left:with stimulation, right:without stimulation)

上下方向の加速度波形の刺激無しと刺激ありの加速度波形の典型例を Fig.4 に示す。

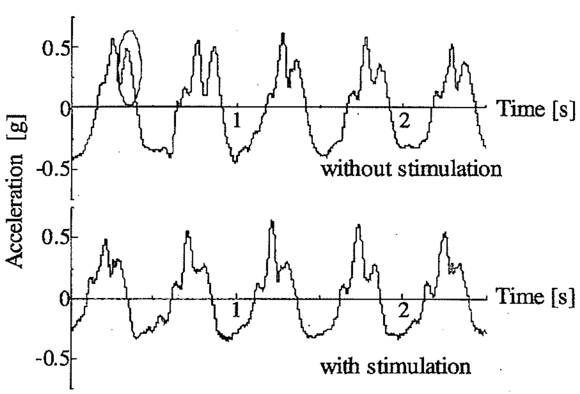

Fig.4　Waveform of vertical acceleration

3. 検討

刺激無しと刺激ありの歩行における前後，左右，上下方向の加速度波形に対して自己相関，rms 値，そして，rms 値から軌跡面積を計算した。

自己相関には両歩行間の差異は見られなかった。

Fig.2 より，被験者 A を除いた他の被験者で刺激ありの場合での加速度3軸全ての rms 値が，刺激無しの場合に比べ減少している。これより，各方向の変位量が小さくなっていると考えられる。Fig.3 より，前後左右方向の加速度の軌跡において，刺激有りの場合の方が軌跡の範囲が全体的に小さくなっている事が確認できる。また，Fig.2 から確認できる rms 値の全体的な減少により，軌跡面積も刺激ありの場合のほうが小さいということが確認された。これより，刺激を与えた場合の方が前後左右方向の歩行中の体の揺れが小さくなっているといえる。また，Fig.4 の上下方向の加速度波形から，刺激ありの場合において波形の 3 つ目のピーク（丸をつけた場所）が小さくなっていることが観察された。これより，蹴り出しの力が弱くなっていると考えられる。

蹴り出しの力が弱くなっているのは，そうすることで遊脚期を短くし，安定期を長く取ろうとしていることが考えられる。しかし，自己相関が両歩行で変化していないため，歩行周期には変化はない。また，上下方向の加速度波形からも周期に変化がないことが分かる。したがって，安定期を長くとろうとする動きではないと思われる。そこで，刺激がある時に蹴り出しの力を弱くすることで，前後，左右，上下各方向の体の揺れを小さくしようとしていると考えられる。これは，足底への刺激を外乱としてとらえ，体を安定させる必要があると認識されるために起こるのではないかと思われる。

以上のことより，本研究で用いた足底刺激装置は，外乱を発生させることで平坦な道でありながら，擬似的な外乱歩行をさせ，体を安定させようとする機能を引き出すと考えられる。そこで，この装置による外乱歩行のトレーニングを長期間行うことで外乱歩行時の体の安定性を高め，通常歩行時におけるバランス機能が改善される可能性が示唆される。今後は，実際に長期間トレーニングを行った際のバランス機能がどのようになるかが検討課題として挙げられる。

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高齢者の歩容に対応した歩数計の開発 ーカウントアルゴリズムの検討ー
Development of a novel pedometer for elderly

関根正樹[1], 木内尚子[1], 前田祐佳[1], 田村俊世[1], 桑江 豊[2], 東 祐二[2], 藤元登四郎[2], 大島秀武[3], 志賀利一[3]
M. Sekine[1], S. Kinouchi[2], Y. Maeda, T. Tamura, Y. Kuwae[2], Y. Higashi[2], T. Fujimoto[2], H. Oshima[3], T. Shiga[3]

[1] 千葉大学工学部, [2] 八日会藤元早鈴病院, [3] オムロンヘルスケア株式会社
[1] Chiba University, [2] Fujimoto Hayasuzu Hospital, [3] Omron Healthcare Co., Ltd

Abstract In this study, we attempted to develop a step-estimation algorithms to improve the accuracy of the accelerometer-type pedometer for elderly people. The waist acceleration signal during walking was recorded from 116 eldery subjects. After three bandpass filters (0.5-10.Hz, 1.0-1.5Hz, 1.5-2.0Hz) were applied to the acceleration, a step-cycle related signal was selected from the outputs of those filters. To estimate the number of steps by the step-cycle signal, optimal threshold level of the signal was determined by using 42 elderly data set. As a result of algorithm verification by 74 elderly data set, 62 subjects (83.7%) could be estimated with less than 10% error.

1. はじめに

生活習慣病の予防や健康増進の1つとして歩行が推奨されている．その量的な評価には，歩数計の利用が一般的であり，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（以下，加速度歩数計）が市販されている．しかし，このような歩数計で，筋力が低下した高齢者や片麻痺などの運動障害をもつ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しばしば見受けられる．

この問題に対して，われわれは得られた加速度を一定期間ごとにフーリエ変換しパワーが最大となる周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名の高齢者に対して誤差10%で歩数をカウントすることが可能であった[1]．しかしながら，常時フーリエ変換を行うことは，歩数計の電池を著しく消耗するため，その実装は容易ではないという問題点が挙げられた．

一般に，加速度歩数計では加速度閾値（以下，閾値）を用いて歩数を算出している．そこで，本研究では先行研究の経験をもとに，簡単なフィルタと閾値処理のみで高齢者に対応しうる歩数カウントアルゴリズムを提案し，その検証を行った．

2. 歩数カウントアルゴリズム

本研究では，加速度信号からフィルタリングと閾値のみで歩数をカウントする以下の 2 つのアルゴリズムを設計した．なお，本アルゴリズムは，測定終了後にオフラインで適用した．また，対象とする加速度の感度方向は上下方向のみとし，上方向を正とした．

2.1 アルゴリズムA

先行研究の結果から，1 歩周期に対応する加速度信号の周波数はおよそ 0.3～3Hz の範囲に含まれる[1]．そこで，歩数カウントアルゴリズムAは，周波数帯域が 0.3～3Hz のバターワース型バンドパスフィルタで加速度信号をフィルタリングし，閾値未満から閾値以上となる点を 1 歩としてカウントするものとした．なお，閾値については後述する実験によって決定した．

2.2 アルゴリズムB

歩数カウントアルゴリズム B は，先行研究のフーリエ変換を用いる手法を簡略化したものである．周波数帯域が異なる3つのバターワース型バンドパスフィルタを設計し，それらを通過した出力のうち最大のものを1歩周期に対応する加速度信号として選択する．この選択された信号が，閾値未満から閾値以上となる点を 1 歩としてカウントした．なお，それぞれのバンドパスフィルタの周波数帯域は，0.5～1.0Hz，1.0～1.5Hz，1.5～2.0Hz とし，出力の比較にはそれぞれの信号を全波整流しカットオフ周波数0.1Hzのバターワース型ローパスフィルタを通したものを利用した．

3. 実験方法

提案した歩数カウントアルゴリズムに必要な閾値の設定ならびにその検証のために以下の実験を行った．なお，本研究は，当該倫理委員会の承認を得た後，被験者に実験の説明し書面にて同意を得て行った．

3.1 測定装置

歩行中に身体に生じる加速度を測定するために，3 軸ピエゾ抵抗型加速度センサ（曙ブレーキ）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た．なお，サンプリング周波数は 128Hz とした．また，比較対象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．

3.2 歩行・歩数の測定方法

被験者は，伸縮性のあるベルトを用いて加速度センサを腰背部中央に装着し，歩数計も同ベルトに取り付けた．歩行は，屋内廊下直線 10m 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，目視にて歩数をカウントした．

3.3 閾値の設定に関する実験

被験者は，転倒予防教室に来場している受講者で脳梗塞を発症したものや骨折経験者，下肢機能障害をもつものを含む高齢者 42 名（年齢 82.2±8.4 歳，身長 146.7±8.5cm，体重 49.5