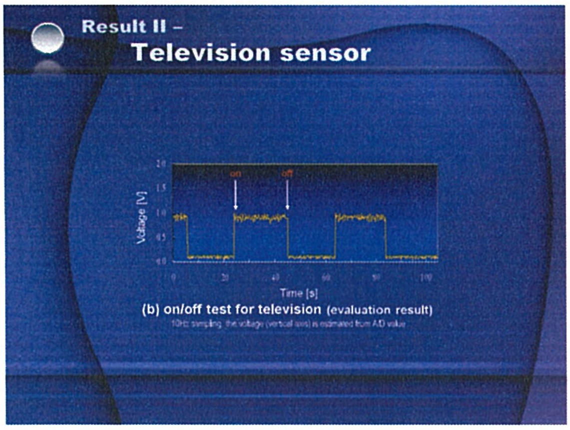
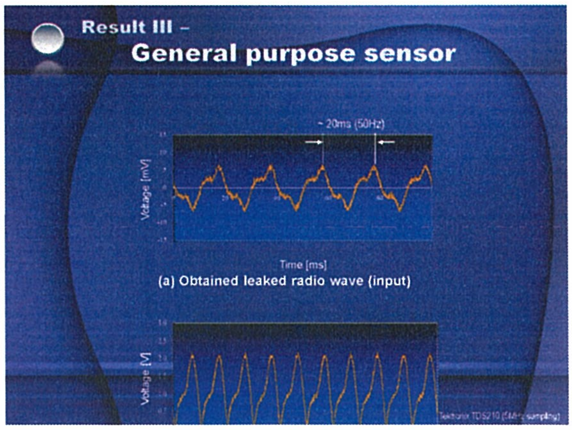
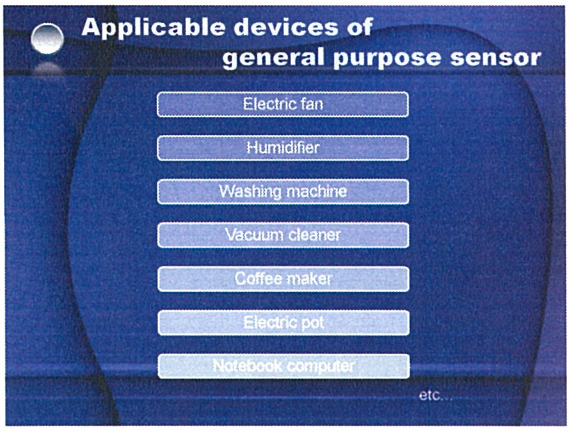
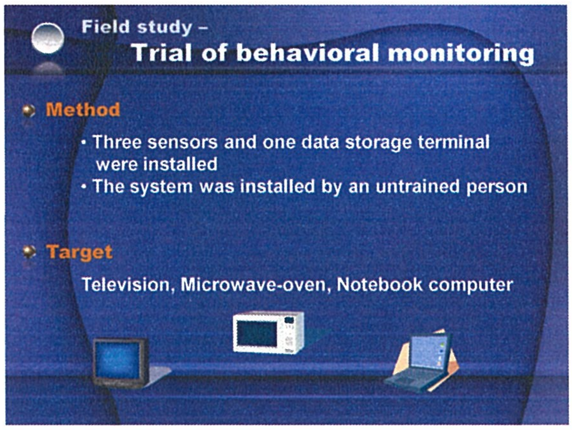
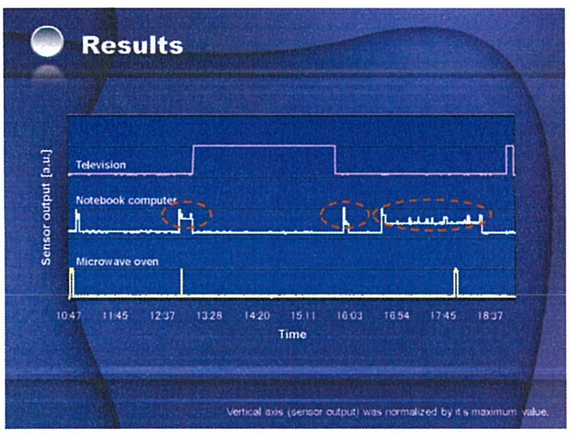
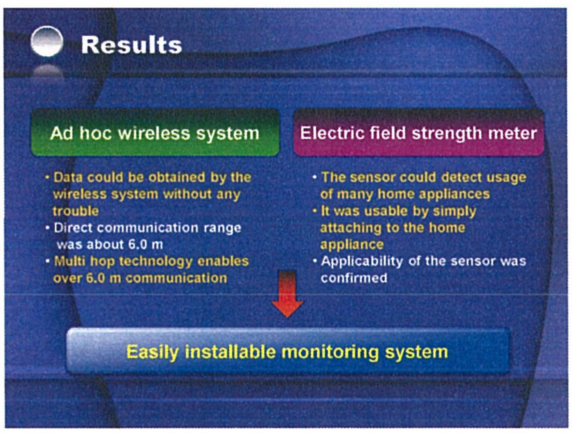

Result II –
Television sensor
Voltage [V]
on
off
Time [s]
(b) on/off test for television (evaluation result)

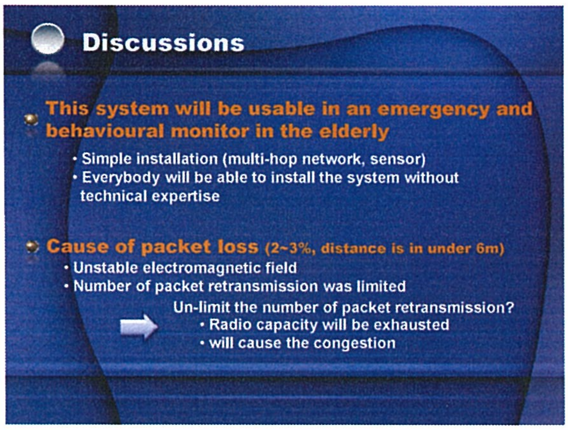

Result III –
General purpose sensor
~ 20ms (50Hz)
Voltage [mV]
Time [ms]
(a) Obtained leaked radio wave (input)
Voltage [V]

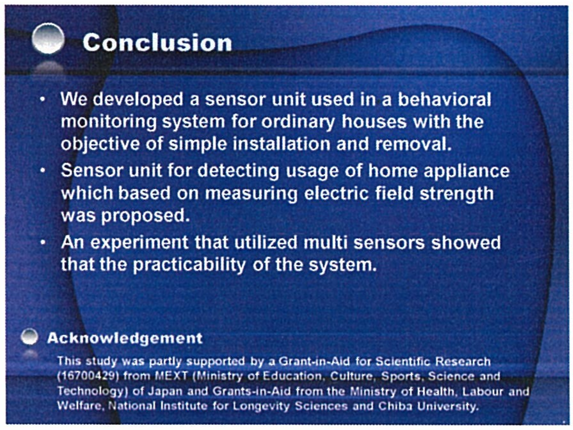

Applicable devices of
general purpose sensor
Electric fan
Humidifier
Washing machine
Vacuum cleaner
Coffee maker
Electric pot
Notebook computer
etc...


Field study –
Trial of behavioral monitoring
Method
• Three sensors and one data storage terminal were installed
• The system was installed by an untrained person
Target
Television, Microwave-oven, Notebook computer


Results
Sensor output [a.u.]
Television
Notebook computer
Microwave oven
10:47 11:45 12:37 13:28 14:20 15:11 16:03 16:54 17:45 18:37
Time
Vertical axis (sensor output) was normalized by it's maximum value.


Results
Ad hoc wireless system
Electric field strength meter
• Data could be obtained by the wireless system without any trouble
• Direct communication range was about 6.0 m
• Multi hop technology enables over 6.0 m communication
• The sensor could detect usage of many home appliances
• It was usable by simply attaching to the home appliance
• Applicability of the sensor was confirmed
Easily installable monitoring system


Discussions
This system will be usable in an emergency and behavioural monitor in the elderly
• Simple installation (multi-hop network, sensor)
• Everybody will be able to install the system without technical expertise
Cause of packet loss (2~3%, distance is in under 6m)
• Unstable electromagnetic field
• Number of packet retransmission was limited
Un-limit the number of packet retransmission?
• Radio capacity will be exhausted
• will cause the congestion


Conclusion
• We developed a sensor unit used in a behavioral monitoring system for ordinary houses with the objective of simple installation and removal.
• Sensor unit for detecting usage of home appliance which based on measuring electric field strength was proposed.
• An experiment that utilized multi sensors showed that the practicability of the system.
Acknowledgement
This study was partly supported by a Grant-in-Aid for Scientific Research (16700429) from MEXT (Ministry of Education, Culture, Sports, Science and Technology) of Japan and Grants-in-Aid from the Ministry of Health, Labour and Welfare, National Institute for Longevity Sciences and Chiba University.

厚生労働科学研究費補助金（医療技術評価総合研究事業）

分担研究報告書

ユビキタスコンピューティングシステムを用いたへき地医療体制の充実に関する研究

分担研究者　南部雅幸　大阪電気通信大学

研究要旨　へき地医療体制充実のためには，情報通信技術を核とした遠隔医療の適用が必要不可欠である．しかしながら，ネットワーク基盤の充実が遅れがちなへき地にあってはしばしば遠隔医療の実現すら困難となる場合がある．本研究では，へき地においても利用可能な携帯電話と，在宅でのネットワークアクセスに有用なユビキタスコンピューティング技術の一部である無線ネットワークを組み合わせ高度な医療を提供するシステムの実現に向け，携帯電話で利用可能なデータベースおよびインターフェースの開発と，生体情報のリアルタイム伝送が可能な通信システムを開発した．

A. 研究目的

へき地医療の充実のためには，遠隔医療の適用が必要不可欠である．ところで，遠隔医療を実施するためにはネットワークのインフラが必要不可欠であるが，基幹ネットワークへのアクセスに関し，個々のユーザに到達するまでのネットワークの整備は，利用者個々の問題となり，均一なサービスを提供するためには，ネットワーク環境の整備が必要である．ところで，携帯電話の普及に伴い，現在では携帯電話４社のサービスを合わせると人口の99.9%が利用可能で，インターフェースを含めれば最も普及したネットワーク端末と考えられる．さらに，ユビキタスコンピューティングシステムの開発推進と，普及の結果，容易に実装可能な無線ネットワークシステムが普及している．本研究では，ネットワークアクセス端末としての携帯電話と，生体情報収集システムとしてのユビキタスコンピューティングシステムを組み合わせた遠隔医療システムを開発し，へき地居住の高齢者等情報弱者にも利用が容易なシステムの開発を行う．

B. 研究方法

本年度は最終年度であるため，より実用化を考慮したシステムの構築を行った．まず，携帯電話を用いた遠隔医療システムを開発し，携帯電話のインターフェース利用を前提とした医療データベースシステムを構築した．携帯電話によるシステムは，

(1)　アニメーションを利用したインターフェースにより容易に情報インフラへのアクセスを可能にするシステム

(2)　日常生活中における生体情報を収集し，蓄積データへのアクセスを携帯電話のみで可能にするシステム

の２つである．

携帯電話の長所である携帯性・ネットワークへのアクセスを考慮し，いつでも，どこでも，ネットワークへのアクセスを可能とすることで，経常的な健康管理を可能とした．また必要に応じて，医療機関，家族な

どへの緊急連絡機能もあわせて実装している．他方，携帯電話の持つ短所である画面の小ささや固定されかつ数字と記号のみの入力装置などの低いユーザビリティを考慮し，日常的に利用することが多い情報のみを抽出・集約しアイコンとアニメーションを用いたインターフェースを再構築した．さらに，キーボードの制限を回避するため，アイコンと数字キーを一意に組み合わせ，家庭用家電機器のリモコンと同等のインターフェースを提供した．
まず，携帯電話利用者と高齢者の希望する情報について，各携帯電話会社および公的機関による公開情報に関する調査と聞き取り調査を行い，必要とされている情報を集約した．さらに，医療情報に関する情報をまとめて追加した．尚，調査の結果，最も利用されている情報のうち，交通機関へのアクセスに関する情報が挙げられたが，へき地医療向けシステムであることを考慮し，これを除外した．これらの情報をもとに，6項目からなるインターフェース画面を作成した．これらの項目はアイコン化されており，選択することで簡単な内容をアニメーション表示することが可能である．以後，それぞれの項目について詳細画面に移動可能であるとともに，すべての項目において数字キーによる選択と記号キーによるコマンド発行に関するインターフェースを統一することで，操作を容易にしている．
同様に，生体情報の入力については，数字の入力と Yes/No 形式の簡単な質問のみでデータの収集が可能である．また結果はサーバに蓄積され，必要に応じて携帯電話のみを用いて数値だけでなく，グラフ化して確認が可能であり，同時に医療機関への提供が可能である．
次に，利用者による任意の入力の問題点を解決するため，無線ネットワークを用いた生体情報収集システムを開発した．同様のシステムはこれまでにも開発・実用化されており，臨床におけるテレメトリシステムはそのひとつである．しかしながら，臨床用テレメトリシステムは，薬事法上の問題がありかつ高価なため日常的に利用することは困難である．そのため，容易に利用可能な Bluetooth を採用した．Bluetooth は，無線通信の規格の一つで，一般的に普及しており，携帯電話，ノート型コンピュータ(以後ノート PC)などにあらかじめ実装されている場合がある．そのため，これらの機器を基地局として利用可能であるため，経常的に計測が可能である．このシステムを人体表面に添付して利用する計測システムに実装することで，利用者の任意性に頼らない計測が可能となる．
（倫理面への配慮）
所属機関の倫理委員会に相当する委員会において研究の安全性に関し審議を行った．実験に先立ち研究の内容に関し口頭で説明を行い文書による承諾を得た．

C. 研究結果
本システムを実際に利用してその有用性を評価した．まず，携帯電話によるインターフェースのユーザビリティの評価を行った．その結果，携帯電話を日常的に利用している利用者で最大 25%のアクセス時間短縮が確認された．尚，高齢者の利用者に限定すると 50%以上のアクセス時間短縮が確認され，一部の被験者では従来システムではまったくアクセスできないケースが存在した．

続いて Bluetooth を用いた生体情報収集システムに関する評価を行った．加速度・筋電センサと Bluetooth を組み合わせ計測を行ったところ，従来のシステムと同様の計測が可能であることを確認した．

D.　考察

生体情報や緊急通報など医療に関する情報を集約して取り扱うことを前提にすると，携帯電話を用いたシステムはWWWをベースにしたシステムが現状最も理想的である．しかしながら，現状の携帯電話用 WWW システムは高齢者などには非常に利用が困難である．一方本システムを利用すれば，アクセスが容易になるため所在地を問わずに必要な情報にアクセス可能であり，へき地においてもほぼ携帯電話の利用が可能な現状を鑑みれば医療のみならず，日常的に利用可能な情報端末として有用であると考えられる．ところで，医療情報は最も重要な個人情報のひとつであるため，情報漏えいに対する防衛策が必要不可欠である．携帯電話は費用対効果の面で理想的ではあるが，セキュリティー機能は別途実装する必要がある．本システムでは，WWW を利用しているため，SSL など既存のセキュリティー機能をそのまま実装可能なため，十分な対策が可能である．

現在ユビキタスコンピューティングシステムにおいて最もよく利用されている無線通信システムは IC カードに代表される RF-ID である．しかしながら，RF-ID は情報のリアルタイム更新や大量の情報伝送が困難である．他方本システムで用いた Bluetooth は，約 1MBps のデータ伝送と巣ペクトラム拡散通信によるセキュリティ機能を有するため，ID 機能，大容量のリアルタイムデータ伝送機能，およびセキュリティー機能をすべて有する点で優位である．ただし，消費電力の点で連続した計測が困難であるが，消費電力の低減と高容量バッテリーの採用で解決が可能である．さらに，携帯電話と組み合わせ情報の収集が可能なため，いつでもどこでも健康診断が可能となる．

[結論]へき地医療用情報システムの一部として携帯電話と無線通信システムを組み合わせた在宅医療システムを開発した．その結果携帯電話のみで十分に利用可能で実用的なシステムが実現できた．特に島嶼部や山間部においても携帯電話の利用可能エリアが拡大しており，携帯電話を用いたシステムはへき地医療の充実に理想的なシステムであると考えられる．

F.　研究発表

１．　論文発表

特になし

２．　学会発表

1. 南部雅幸，田村俊世，"Bluetooth を用いた生体情報モニタリングシステム"，計測自動制御学会システム・情報部門学術講演会 2006，東京，2006

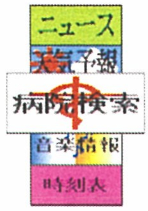
2. 南部雅幸，田村俊世，“Bluetooth を用いた生体情報モニタ”，日本生体医工学会「在宅医療と ME 技術」研究会，神戸，2006

G.　知的財産権の出願・登録状況

特になし

ユビキタスコンピューティングシステム
を用いた
へき地医療体制の充実に関する研究

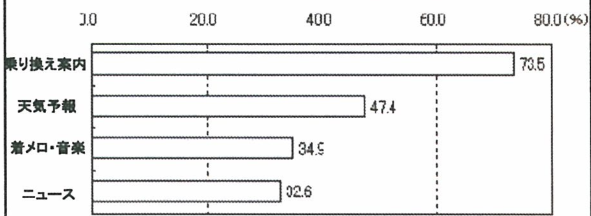
アイコンによる携帯電話向けメニュー



病院検索のページ
（選択されたメニューの項目を表示）

添付資料

最もよく利用される情報



Q. どんなWEBサイトにアクセスすることが多いですか。(複数回答)
<ベース：インターネット機能使用者 (n = 215)>

公開調査データより

添付資料

医療情報探索機能

数字キーで選択

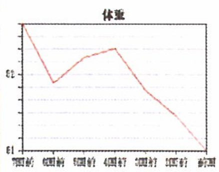
病院検索を行うと連絡先，位置，地図を表示

添付資料

日常的生体情報収集システム

入力画面

グラフでの表示



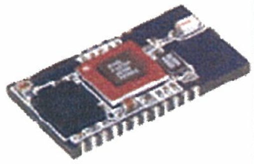
添付資料

Bluetoothの特長

- 短距離だが高速な通信が可能(1Mbps)
- 様々な周辺機器を接続可能
- 自動識別
- 暗号化通信

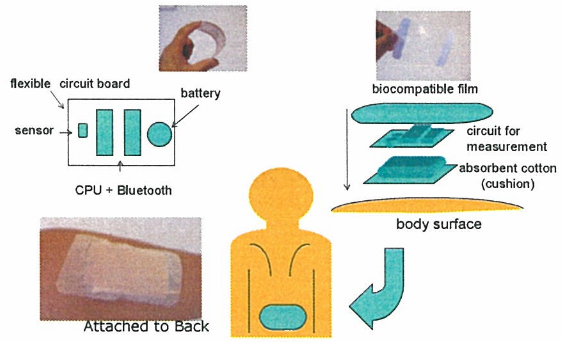
添付資料

Bluetoothチップの例

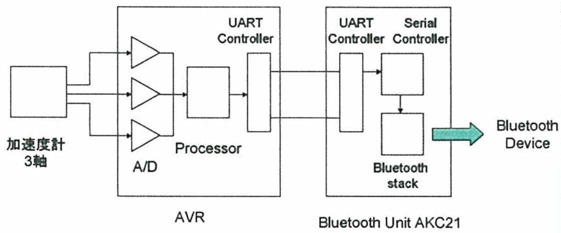


- ラビットセミコンダクターAKC21
- Class2
  出力2.5mW Max30m (10m保証)
- UART 入力可
- 15mm x 27mm x 4mm
- 3V 駆動
- アンテナ オンボード
- プロトコルスタック内蔵
  SPPモードで利用することで，仮想シリアルポートとして利用可能.

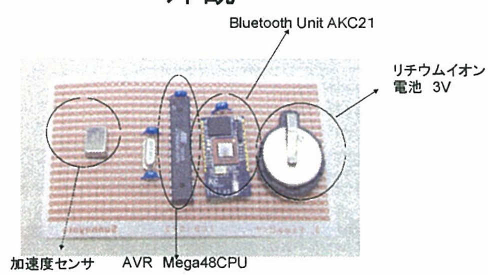
体表面添付型計測システム



システム構成



外観



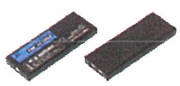
105mm x 50mm x 7mm（厚み）

電力消費について

- 消費電流試算 3V 駆動時 （カタログ値）
  - CPU: 1mA
  - Bluetooth unit: 35mA 通信時 (ピーク時100mA)
  - 加速度センサ: 1.5mA

  合計 40mA 以下
- 電源CR2032リチウムイオンコイン電池
  - 容量:220mAh → 5時間の駆動可能

- 実際の計測時における利用可能時間
  - 10回の試行で2-3時間
- 現状では常時計測は不可能である.



- 携帯電話用二次電池の採用を検討
- 省電力用制御の導入
- 運用方法の検討 → 一日一回取り替え／リユース

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

刊行書籍又は雑誌名 (雑誌のときは雑誌名　巻○○頁～○○頁　論文名)	刊行年	刊行者氏名	執筆者氏名
Medicine & Science in Sports & Exercise. 38(9):1674-81, Improving the accuracy of pedometer used by the elderly with the FFT algorithm.	2006	American College of Sports Medicine	Ichinoseki-Sekine N, Kuwae Y, Higashi Y, Fujimoto T, Sekine M, Tamura T
Journal of Telemedicine and Telecare. 12;203-207, Low-Cost, email-based system for self blood pressure monitoring at home.	2006	RSM Press	Nakajima K, Nambu M, Kiryu T, Tamura T, Sasaki K
Assistive Technology Research Series 19: SMART HOMES AND BEYOND. 2006:7-11, A Smart House for Emergencies in the Elderly.	2006	IOS Press	Tamura T
Assistive Technology Research Series 19: SMART HOMES AND BEYOND. 2006:212-219, Easily Installable Sensor Unit Based on Measuring Radio Wave Lealage from Home Appliances for Behavioural Monitoring.	2006	IOS Press	Tsukamoto S, Akabane Y, Kameda N, Hoshino H, Tamura T
Proc. of 28th Annual International Conference of IEEE Engineering in Medicine and Biology Society:6261-6264 (CD-ROM), Proposal of Wireless Behavioral Monitoring System with Electric Field Sensor.	2006	IEEE	Kameda N, Akabane Y, Naganokawa H, Tsukamoto S, Tamura T, Hoshino H
第21回生体・生理工学シンポジウム講演予稿集:3A1-3, 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	2006	BPES	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

SICE Symposium on Systems and Information 2006 講演予稿集:21-26, 電界強度計による電化製品稼動状況モニタリングシステム.	2006	SICE	塚本壮輔, 亀田倫之, 町田雄一郎, 田村俊世, 星野　洋
SICE Symposium on Systems and Information 2006 講演予稿集:35-38, Bluetoothを用いた生体情報モニタリングシステム.	2006	SICE	南部雅幸, 田村俊世

IV.　研究成果の刊行物・別刷

SPECIAL COMMUNICATIONS
Methodological Advances

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]

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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

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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

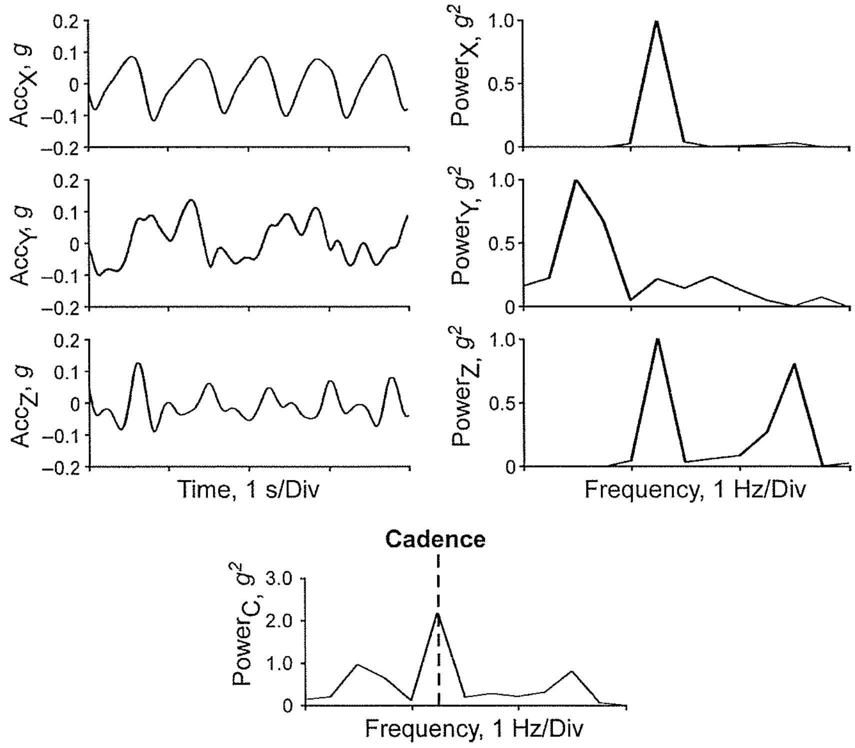

FIGURE 1—Typical examples of the acceleration signals, Acc_X, Acc_Y, Acc_Z, and their normalized power spectrum, $power_X$, $power_Y$, and $power_Z$. The frequency at the maximum power of the composite power spectrum, $power_C$, was considered as the cadence of each window.

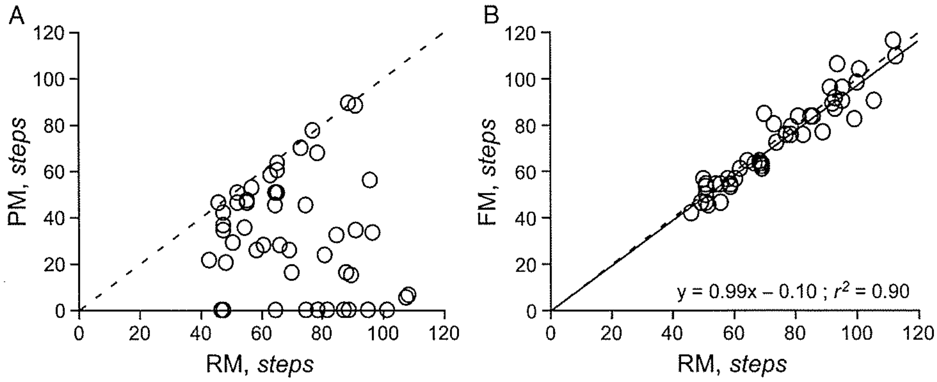

FIGURE 2—Relationship between number of steps counted visually by a physical therapist (RM) and A) number of steps counted by pedometer (PM) and B) number of steps estimated using the FFT algorithm (FM). FM significantly correlated with the actual number of steps ($P < 0.0001$).

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.

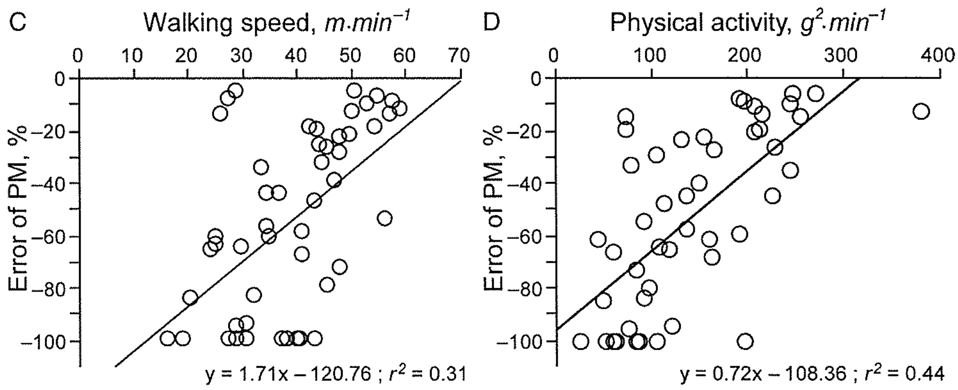

FIGURE 3—Relationship between the error of PM (% RM) and C) walking speed and D) physical activity in the vertical direction. The error of PM was significantly correlated with the walking speed and physical activity (P < 0.0001).

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

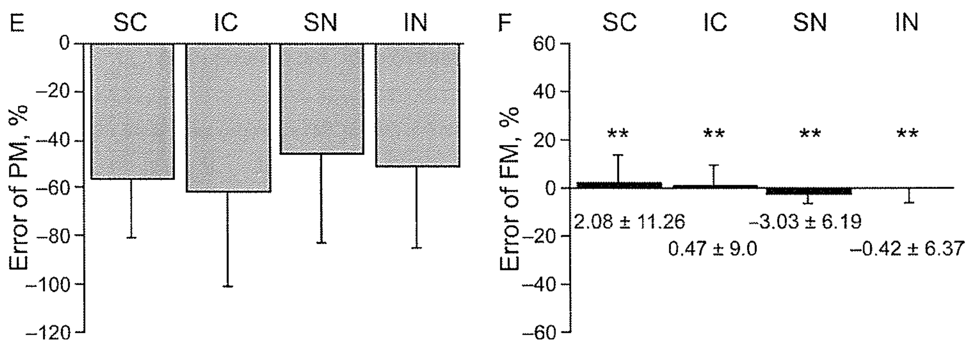

FIGURE 4—The errors of PM (% RM) (E) and the error of FM (F) for senile gait with cane (SC), impaired gait with cane (IC), senile gait without cane (SN), and impaired gait without cane (IN). ** P < 0.01 vs the errors of PM within the same group.

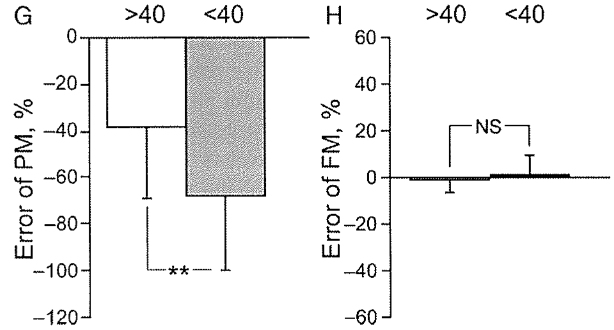

FIGURE 5—Effect of speed on the error of PM (% RM) (G) and the error of FM (H). > 40, greater than 40 m·min^{-1} average walking speed group, N = 26; < 40, less than 40 m·min^{-1} average walking speed group, N = 23. ** P < 0.01; NS: not significant.

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

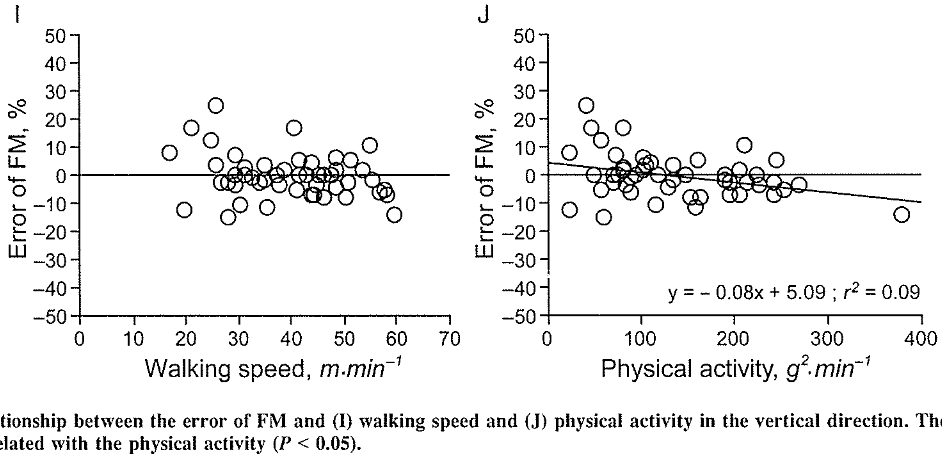

FIGURE 6—Relationship between the error of FM and (I) walking speed and (J) physical activity in the vertical direction. The error of FM was significantly correlated with the physical activity (P < 0.05).

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 overestimated 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

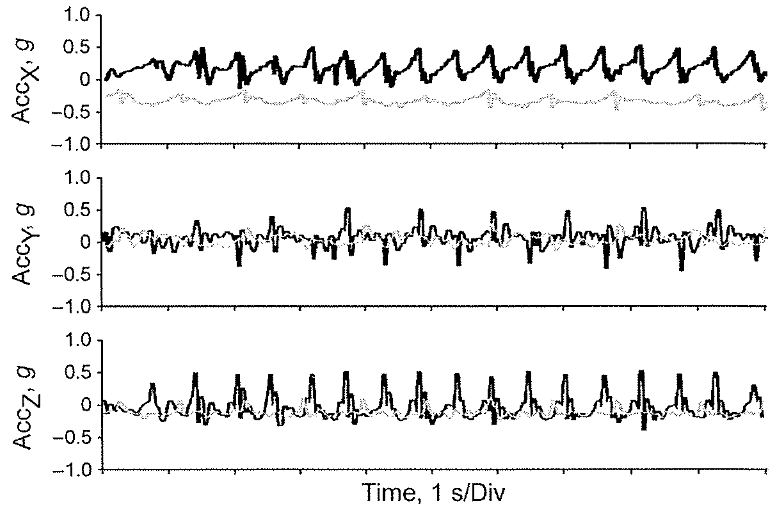

FIGURE 7—Typical examples of the acceleration signals Acc_X, Acc_Y, and Acc_Z in well-counted subjects (PM was more than 90% of RM; *black line*) and uncounted subjects (PM was 0% of RM; *gray line*).

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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Low-cost, email-based system for self blood pressure monitoring at home

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Summary

We have developed a low-cost monitoring system, which allows subjects to send blood pressure (BP) data obtained at home to health-care professionals by email. The system consists of a wrist BP monitor and a personal computer (PC) with an Internet connection. The wrist BP monitor includes an advanced positioning sensor to verify that the wrist is placed properly at heart level. Subjects at home can self-measure their BP every day, automatically transfer the BP data to their PC each week, and then send a comma-separated values (CSV) file to their health-care professional by email. In a feasibility study, 10 subjects used the system for a mean period of 207 days (SD 149). The mean percent achievement of measurement in the 10 subjects was 84% (SD 12). There was a seasonal variation in systolic and diastolic BP, which was inversely correlated with temperature. Eight of the 10 subjects evaluated the system favourably. The results of the present study demonstrate the feasibility of our email-based system for self-monitoring of blood pressure. Its low cost means that it may have widespread application in future home telecare studies.

Introduction

Self blood pressure monitoring (SBPM) at home is important for patients with hypertension in order to teach them about their chronic blood pressure (BP) condition and to allow a more efficient assessment of the patient's response to antihypertensive medication. SBPM also removes the pressor effect of the office, eliminates observer errors and enhances patient compliance, potentially reducing costs and providing objective confirmation of readings taken by patients or elderly people.[1] In addition, previous studies have demonstrated that SBPM has a stronger predictive power for target organ damage, morbidity and mortality than that of conventional office BP measurements.[2]

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Electronic, oscillometric BP monitors, which are compact and easy to use (even for elderly people) have been developed for SBPM.[3–5] These monitors use the upper arm,[3] wrist[4] or finger[5] as the measurement location. However, it has been noted that finger measurement is not suitable for accurate BP monitoring.[6,7] In addition, wrist cuffs are more comfortable than upper arm cuffs for repeated BP measurement.[8,9] The accuracy of wrist measurements depends on proper positioning of the wrist at heart level. Because of the uncertainty of the wrist level during measurements, some professionals are reluctant to recommend wrist BP monitors.[9,10] To overcome this problem, wrist BP monitors with a positioning sensor have been developed recently and are now commercially available.[9,11] These may be valuable for SBPM.

We have developed a low-cost email-based system, which allows patients and elderly people to send BP data obtained at home to health-care professionals such as a general practitioner at regular intervals, and to receive advice from them on a regular basis. The continuous dialogue between the health-care