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A Network-Based Monitoring System for Rehabilitation

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Abstract—**Various techniques have been developed to monitor a patient's motion for rehabilitation, but many of them are expensive or inadequate for monitoring motions in daily living. In this paper, a network-based monitoring system, which consists of wireless sensor modules, computers at local and remote sites connected via the Internet, and a user-friendly monitoring program, is proposed. The wireless sensor module is an array of lightweight Arduino-powered inertia measurement units (IMUs) with wireless communication powered by the ZigBee protocol. The wireless sensor modules measure kinematic information of a human body, and the measured data is analyzed at the local and host computers which are connected via the Internet. For easy observation and monitoring, a 3D human model is animated in both computers. In order to verify the performance of the proposed system, gait motions are monitored and analyzed by the proposed system with a shoe-type ground reaction force measurement system. The experimental results show that the proposed monitoring system can be used as a method to cheaply and noninvasively provide kinematic information conducive towards rehabilitation for patients, even without the presence of a trained specialist.**

I. Introduction

As the number of patients or elderly people who need rehabilitation treatments increases [1-3], the demand for rehabilitation therapy also increases. Since observation and evaluation of the patient's status is one of the most important and fundamental processes in rehabilitation, assessment methods based on various functional tests such as Fugl-Meyer [3], Functional Independence Measure (FIM) [5], and Barthel Index [6] have been widely used to assess motor and sensory impairment. However, observation and assessment of dynamic motions in daily living such as walking motions is still quite difficult, thus it mainly depends on physical therapists' experience, knowledge, and observation skill. In order to quantitatively measure and assess such motions, the required sensor sets are usually complicated and applicable only in limited places such as research laboratories or hospitals. Also, rehabilitation treatments are applied to patients only when they visit a rehabilitation facility, thus the rehabilitation effect only lasts during the rehabilitation sessions. In order to overcome the aforementioned problems, a network-based monitoring system, which takes advantages of an inertia measurement unit (IMU), local wireless network such as ZigBee, the Internet, and a user-friendly monitoring program, is proposed in this paper.

As a motion measurement system, camera-based systems such as VICON [7] have been widely used. In this method, several optical markers are mounted on a human body, and infrared cameras capture the reflected light from the markers. It produces well-quantified and accurate results on joint motions, thus it is known as a gold standard method to measure kinematic and kinetic motion data. However, the camera-based method is expensive and restricted to a laboratory environment.

Sensors can be directly attached to a human body to measure motion information. Bio-sensors such as an electromyography (EMG) sensor have been used to measure and estimate human joint torques or posture [8-10], but the wide use of the EMG sensor is limited by its sensitivity and required peripherals. Motion sensors such as an encoder or an accelerometer can be attached to a human body with an exoskeleton-type linkage [11,12], but wearing such devices may disturb the user's motions due to insufficient degrees of freedom of the linkage. Also, the exoskeleton-type linkage can be used to measure motions only in the sagittal plane.

In this paper, a network-based monitoring system for rehabilitation, which uses IMUs, local wireless network by ZigBee, the Internet, and user-friendly monitoring program, is proposed to allow unobtrusive and remote monitoring. The wireless sensor modules, which consist of a ZigBee transmitter, an IMU, and a microprocessor, are attached on body segments where the motion information is measured. The wireless sensor modules communicate with the coordinator node in the local computer, which then transmits the measured and estimated information to the physical therapist's host computer via the Internet. By using the Internet, a tele-monitoring is achieved. A 3D model of a human body is animated in both computers at local and remote sites in real-time by the transmitted data.

This paper is organized as follows. In section II, the previous work about the monitoring system is introduced with its achievement and limitation, which is the motivation of this paper. The configuration of the proposed network-based monitoring system is introduced in section III. The detailed monitoring algorithms by the wireless sensor modules and the monitoring program interface using a 3D human model are presented in section IV. The proposed system and algorithms were applied for gait analysis, and the experimental results are discussed in section V. Conclusion and future work are given in section VI.

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Fig. 1. A mobile gait monitoring system [13]

II. Previous Work: A Mobile Gait Monitoring System

In the previous work [13], a mobile gait monitoring system (MGMS) was proposed for the diagnosis of abnormal gait and rehabilitation. The MGMS consists of a shoe-type ground reaction force (GRF) measurement system called Smart Shoe and a micro-processor with a touch screen display, as shown in Fig. 1. The Smart Shoe shown in left of Fig. 1 embeds novel force sensors, each consisting of an air bladder made by winding a silicone tube and an air pressure sensor, under the insole. When the foot presses the air bladder, the pressure change inside of the air bladder is measured by the air pressure sensor, and the measured air pressure is converted to GRF. To capture the foot pressure distribution during walking, four air bladder sensors are installed in Smart Shoes at the hallux, the first metatarsophalangeal joint, the fourth metatarsophalangeal joint and the heel.

The MGMS monitored patients' gait by observing GRF and the center of GRF, and analyzed the gait abnormality. Since visual feedback about patients' GRFs and normal GRF patterns were provided by the MGMS, patients could practice the rehabilitation treatment by trying to follow the normal GRF patterns. Also, the gait abnormality, which was defined by the deviation between the patient's GRFs and normal GRF patterns, was used to quantify the degree of abnormality of the patient. The small size and light weight of the MGMS enabled mobile gait monitoring. For the detailed information of the Smart Shoes or the MGMS, see [13,14].

The effectiveness of the MGMS has been verified by clinical tests with patients suffering from gait disorders, but the limited information provided by the MGMS, i.e. the GRF information, was not enough to show the abnormal gait for some patients. For example, a Parkinson's disease patient, who swung his leg away from the body due to a stiff knee joint showed almost normal GRF signals, but the gait motion was obviously abnormal. Also, the patient could get the feedback from a physical therapist only when they were in the same place together. To overcome such problems, in this paper, direction information is estimated using IMUs, and it is wirelessly transmitted using the local wireless network and the Internet for unobtrusive and remote monitoring.

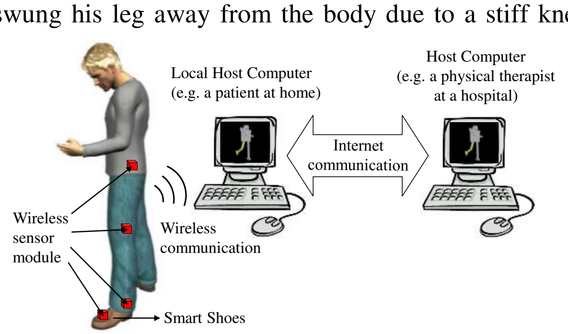


Fig. 2. Concept of the network-based monitoring system for rehabilitation

III. Configuration of the Network-Based Monitoring System for Rehabilitation

A. General Description

The proposed network-based monitoring system consists of wireless sensor nodes which are attached on a human body, computers at the patient's and the physical therapist's sides, and a user-friendly monitoring program. The sensor nodes are placed at strategic locations on a user's body (e.g. knee, ankle, and foot) and send kinematic information to the coordinator node of the local host computer. The wireless sensor modules and the coordinator node communicate with each other via the ZigBee protocol. The coordinator node relays the collected data via serial line to the local host computer. The analyzed data is transmitted to another host computer at a remote site (i.e. a rehabilitation facility) via the Internet. Both the local and remote host computers process the data to analyze motion information as well as to provide a user-friendly graphical interface, which includes a 3D human animation. The concept of the proposed network-based monitoring system is shown in Fig. 2.

B. Local Wireless Communication Setup

For the measurement of kinematic information of a human motion, a camera-based method [7] or wired sensors [11,12] have been widely used. However, the camera-based method is limited only in a laboratory setting, and the wired sensors on a human body may interfere the motion by additional resistance and a limited motion range. The proposed sensor node in this paper addresses these concerns by virtue of its design and hardware.

1) *Sensor Node and Coordinator Node*: In the design of the wireless sensor nodes, attention was paid towards minimizing the 'presence' of sensor nodes on users. Each sensor node is a lightweight, non-intrusive device composed of: 1) a custom-printed circuit board, 2) a 9 DOF inertia measurement unit (ADXL345 3D accelerometer, an ITG3200 3D gyroscope, and a HMC5883L 3D magnetometer) [15], 3) a 16MHz Arduino pro microprocessor [16], 4) a 1mW ZigBee Series 1 Wire Antennae chip [17], and 5) a 7.4V Li-Po battery. The size of a sensor node is 30×20×20 mm and weight is about 10 g including a battery. The wireless sensor node is shown in Fig. 3.

The coordinator node is a ZigBee Series 1 radio module connected via a USB breakout board to a computer. It serves as a link between the computer and the radio network; sending query packets over the radio network and relaying the received information through the serial line to the computer.

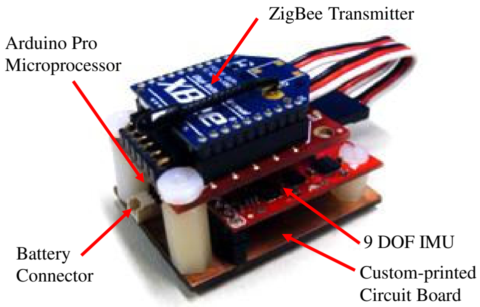


Fig. 3. Wireless sensor node

2) *Communication Protocol*: The proposed local wireless network uses the IEEE 805.15.4 standard for packet transmission from a sensor node to a coordinator node. The IEEE 805.15.4 standard is built in the ZigBee chips and allows setting up local wireless networks with minimal infrastructure while consuming small amounts of power. Built on top of this communication protocol is the proprietary ZigBee packet configuration, which allows packets to be addressed to particular nodes and innately provides a level of packet loss prevention.

Fig. 4 shows the data flow of the sensor node and the coordinator node. The measured sensor data from the IMU are processed in the Arduino chip. For data transmission using ZigBee, appropriate header is appended to the data at the sensor node and coordinator node. The coordinator node sends out a query packet addressed to a remote sensor node to prompt the sensor node to respond with data. The wireless sensor nodes send an acknowledgement packet when a query packet is received.

C. *Multipoint Communication*

In a wireless body sensor network, multiple sensor nodes are used to measure salient information from different locations (e.g. information at the knee, ankle, and foot) which is then integrated together to produce useful data for analysis. Several different procedures for accessing information from multiple nodes have been tried.

One of the first methods attempted was gathering data by querying a sensor node and then waiting for it to respond before moving on to the next node. The step was then repeated with the next node in the network until all nodes are queried. At this point, the coordinator node addresses the first node again and the cycle repeats as shown in Fig. 5(a). Using this method, packet fidelity was high and packets were rarely mixed up (i.e. sensor node 1 data interpreted to be sensor node 2 data). However, the fidelity came at the expense of data transmission rate, which was found to be around 10Hz for complete sampling of three sensor nodes, which is too slow to monitor a human motion.

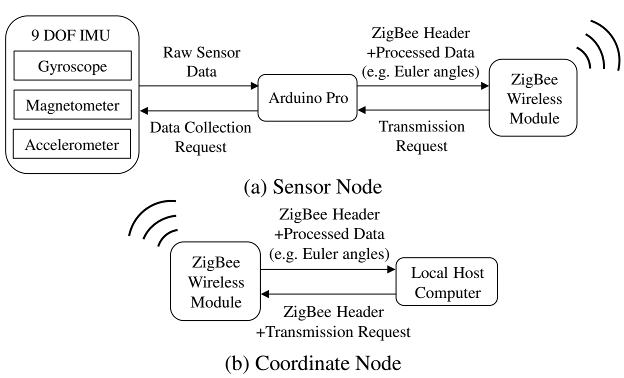


Fig. 4. Wireless communication process

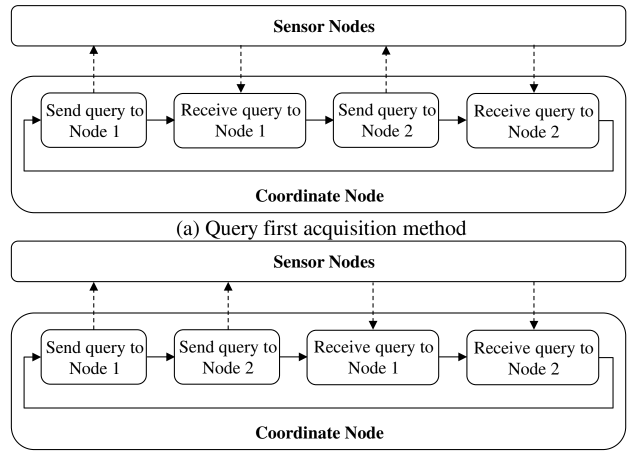


Fig. 5. Multipoint communication methods

Another data acquisition method attempted was multiplexing the coordinator node to send out a series of query packets to all sensor nodes and then collecting the response packets from the sensor nodes as shown in Fig. 5(b). This method relies on the ZigBee's inbuilt clear channel assessment, which ensures that activity on the channel falls below an adjustable threshold before a packet can be transmitted. This assessment prevents multiple nodes from responding to the coordinator's data requests simultaneously. If the assessment fails, the ZigBee waits and re-perform the assessment again before reattempting to send a packet. This acquisition method allows the coordinator node to have a much higher duty cycle as it does not have to wait for each sensor node to process the query, assemble a data packet, and transmit before sending another query. Under this data acquisition scheme, the average transmission rate was 25Hz for a wireless sensor network with three nodes.

D. *Compensation Algorithms for Dropped Packets*

Since radio communication is subject to interference, any protocol over a radio connection must account for the possibility of lost or corrupted data. To account for these potential problems, the ZigBee firmware appends a one byte checksum to the end of every transmitted packet. When a recipient node receives a packet, it uses the checksum to verify the received packet. If the checksum does not match the packet, the recipient node disregards the packet and does not return an acknowledgement packet to the original packet sender. If the sending node does not receive an acknowledgement packet after transmission, it treats the packet as lost and retries to send the same packet until the transmission is successful or all retry attempts are exhausted. If retry attempts are exhausted without a successful transmission, the packet is abandoned. The serial line connection to the computer does not show this resend process at all, thus the only effect from a successful corrupted packet compensation would be the little delay (< 1 ms) to resend the data. In the case that a packet is not successfully transmitted from a sensor node, the coordinator node returns an NaN to the local computer, which then interprets it as an empty string. At this point, the source code written in LabVIEW in the computer replaces this missing packet with the previous packet stored in memory (i.e. if the k^{th} packet is absent, it is replaced with the $(k\text{-}1)^{th}$ packet) because scope of usage for the proposed sensor network deals only with small changes over time.

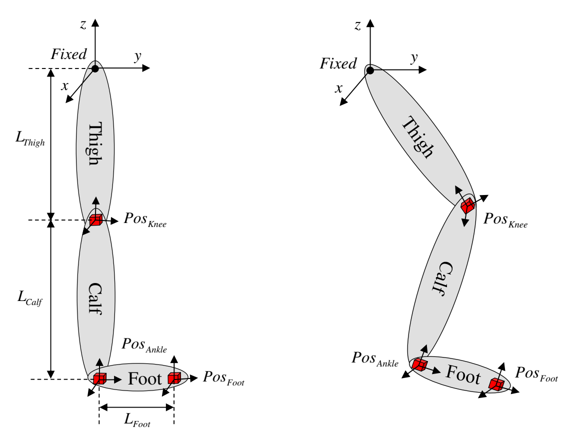
E. Internet Communication

For the tele-monitoring via the Internet, TCP (transmission control protocol) is used as in the usual Internet communication. TCP is known as a more reliable data transmission protocol than UDP (user datagram protocol) because TCP uses an acknowledgment scheme to verify that the signal is correctly delivered [19]. The packet is resent if the sender does not receive an acknowledgement packet before the specified time. To use the resent packet without loss, packet buffers with appropriate size are required in the Internet communication programs. In the actual experiments, the Internet communication was achieved by LabVIEW with appropriate packet buffer size in each computer.

IV. Monitoring Algorithms and Program Interface

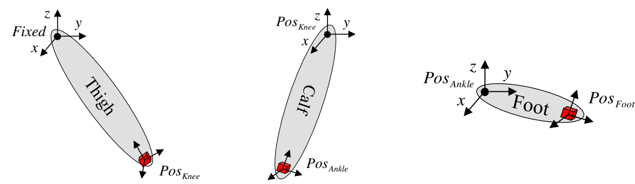
A. Estimation of Position with a Human Model

Orientation estimation is done on the Arduino chip with the ‘Direction-Cosine-Matrix’(DCM) method [20]. This method is specifically developed to work with low cost IMUs that have high noise and time varying biases. It is also tailored to work with the memory constraints of low power embedded systems. The DCM method uses direct calculation on the gyroscope data to produce an estimate of the rotation matrix, and then refines the yaw value with magnetometer data and the pitch and roll values with accelerometer data. From the rotation matrix, Euler angles (α, β, γ) with respect to x, y, and z axis, are extracted and transmitted.



(a) Initial position (b) Moving position

Fig. 6. A leg model with the attached wireless sensor modules (in red)



(a) Thigh (b) Calf (c) Foot

Fig. 7. Body segments with wireless sensor modules

The collected data from each sensor node are integrated to analyze the human motion using a human model. In this paper, a leg model is applied as an example for motion analysis. Although a leg model is applied in this paper, similar analysis can be extended to other body segments. Suppose three wireless sensor modules are attached at the end of the thigh, the calf and the foot to estimate the leg motion as shown in Fig. 6. Since the lengths of each body segment, L_{Thigh}, L_{Calf} and L_{Foot}, can be easily measurable, they are assumed to be known. The hip joint, the origin at the top in the figure, slightly moves up and down during walking, but it is assumed to be fixed in this model.

Fig. 7 shows the body segment with wireless modules. The positions of each wireless sensor module, Pos_{Knee}, Pos_{Ankle} and Pos_{Foot}, are calculated as follows:

$$Pos_{Knee} = R_{Knee} \cdot Pos_{Knee,Init} \tag{1}$$

$$Pos_{Ankle} = P_{Knee} + R_{Ankle} \cdot Pos_{Ankle,Init} \tag{2}$$

$$Pos_{Foot} = P_{Ankle} + R_{Foot} \cdot Pos_{Foot,Init} \tag{3}$$

where the initial positions of each sensor module, $Pos_{Knee,\ Init}$, $Pos_{Ankle,\ Init}$ and $Pos_{Foot,\ Init}$, are defined as,

$$Pos_{Knee,Init} = (0,0,-L_{Thigh}) \tag{4}$$

$$Pos_{Ankle,Init} = (0,0,-L_{Calf}) \tag{5}$$

$$Pos_{Foot,Init} = (0, L_{Foot}, 0) \tag{6}$$

and the rotation matrices of each sensor module, R_{Knee}, R_{Ankle} and R_{Foot}, are,

$$R_{Knee} = R(\alpha_{Knee}, \beta_{Knee}, \gamma_{Knee}) \tag{7}$$

$$R_{Ankle} = R(\alpha_{Ankle}, \beta_{Ankle}, \gamma_{Ankle}) \tag{8}$$

$$R_{Foot} = R(\alpha_{Foot}, \beta_{Foot}, \gamma_{Foot}) \tag{9}$$

The rotation $R(\alpha, \beta, \gamma)$ with Euler angles of (α, β, γ) is calculated as follows:

$$R(\alpha,\beta,\gamma) = R_x(\alpha) \cdot R_y(\beta) \cdot R_z(\gamma) \tag{10}$$

where

$$R_z(\alpha) = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos\alpha & -\sin\alpha \\ 0 & \sin\alpha & \cos\alpha \end{bmatrix} \tag{11}$$

$$R_y(\beta) = \begin{bmatrix} \cos\beta & 0 & \sin\beta \\ 0 & 1 & 0 \\ -\sin\beta & 0 & \cos\beta \end{bmatrix} \quad (12)$$

$$R_x(\gamma) = \begin{bmatrix} \cos\gamma & -\sin\gamma & 0 \\ \sin\gamma & \cos\gamma & 0 \\ 0 & 0 & 1 \end{bmatrix} \quad (13)$$

B. Monitoring Program Interface

The position data returned from the sensor nodes can be interpreted in a variety of ways to assist observation and diagnosis of the patient. The 3D graphical model of a human body in Fig. 8 reflects the positions transmitted by the sensor nodes. The human model is animated in real-time, and five different views (left, right, front, rear and perspective views) are provided in both computers at the patient's and the physical therapist's sides. In this paper, the GRF information measured by Smart Shoes is also displayed at the bottom for the application of gait analysis.

V. EXPERIMENTAL VERIFICATION

A. Performance Verification of the Wireless Module

The performance of the proposed monitoring system mainly relies on the sensing ability of the wireless modules. In order to verify the performance of the wireless modules, a simple linkage was made with two joints as shown in Fig. 9. Encoders were attached to each joint to measure the joint angle, which were assumed to be true values; the wireless sensor modules were attached to the link as shown in the figure. The top part of the linkage was fixed while the two subsequent links were moved back and forth. Both the encoders and the wireless sensor modules measured the joint angles at the same time. Since the encoders measure the relative angles from the initial position, which was a vertically aligned position, the relationship between the encoder measurement and the wireless sensor module measurement is as follows:

$$\theta_{Hip,Enc} = \theta_{Hip,IMU} \quad (14)$$

$$\theta_{Knee,Enc} = \theta_{Hip,IMU} + \theta_{Knee,IMU} \quad (15)$$

Fig. 8. Monitoring program interface

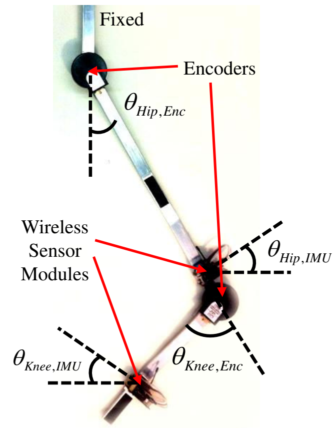


Fig. 9. Experiment setup for verification of the wireless modules

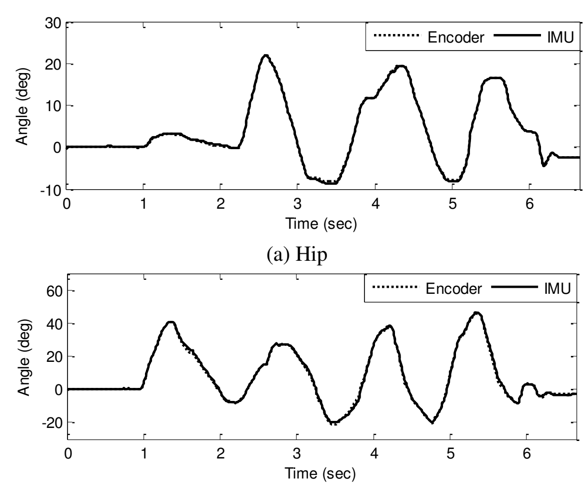
where the encoder measurement, $\theta_{Hip,Enc}$ and $\theta_{Knee,Enc}$, and the wireless sensor module measurement, $\theta_{Hip,IMU}$ and $\theta_{Knee,IMU}$, are shown in Fig. 9.

The experimental results are shown in Fig. 10. The graphs compare the relative angles measured by the encoders and the wireless sensor modules using (14) and (15). As shown in the figure, the wireless IMU modules' estimated angles match the true angles measured by encoders with small error.

B. Gait Analysis: Normal Walking

The proposed wireless sensor modules and algorithms were applied for gait analysis. As shown in Fig. 11, three wireless sensor modules were attached at the end of the thigh, the calf and the foot, respectively, with Velcro, and Smart Shoe was also applied to measure GRFs at the same time. Four healthy persons (three males, one female, age: 23.75±4.32) without any known gait disorders have been participated in the walking experiment. The participants were asked to walk on the plain treadmill about 30~50 m at 4 km/h, which is in a normal walking speed range. Ten consecutive steps data from each participant were collected, and the mean and standard deviation were calculated.

The experimental results in Fig. 12 show estimated joint angles in 3D space. The experimental results show that joint motions on the frontal and transverse planes are much smaller



(a) Hip

(b) Knee

Fig. 10. Performance verification of the wireless sensor module

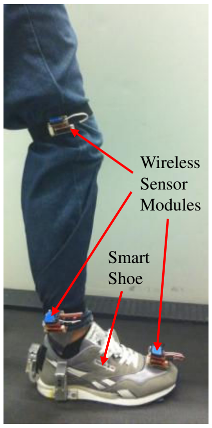


Fig. 11. Experimental setup for gait analysis

than those on the sagittal plane. After 60% of stride, the ankle and the knee show relatively large motion in the frontal and transverse planes due to the swing motion, but the main gait motions are on the sagittal plane. The measurements in the sagittal plane were verified by comparing with references [21].

VI. Conclusion

In this paper, the network-based monitoring system for rehabilitation was proposed. The proposed system consists of wireless sensor modules, a local host computer at the patient's side, and a host computer at the physical therapist's side. The wireless sensor modules estimate orientation information using the embedded inertia measurement unit (IMU), and wirelessly communicate with the local host computer through the ZigBee protocol. The collected data from several wireless modules are aggregated and analyzed in the local host computer and then transmitted to the host computer via the Internet. For easy monitoring, a human model is animated with the estimated orientation data in both computers. The proposed system and algorithm have been applied for gait analysis with a shoe-type ground reaction force (GRF) measurement system called Smart Shoe, and the results showed that the gait motions can be easily observed by simply attaching the wireless sensor modules on the desired body segments.

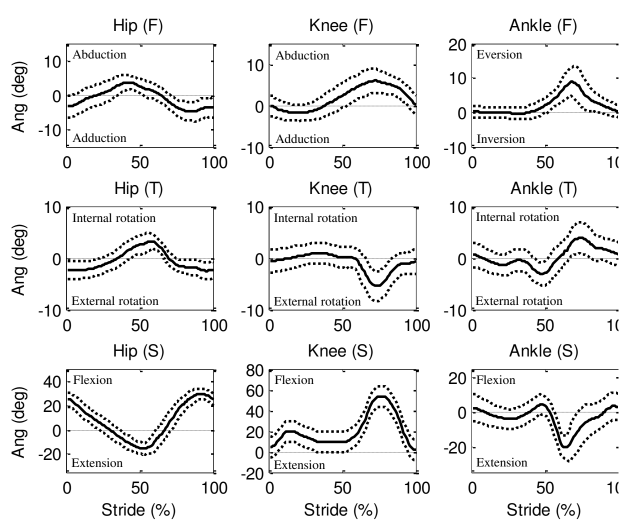


Fig. 12. Experimental results: estimation of joint angles in 3D (solid lines: mean, dotted lines: ±1.96SD, F: frontal plane, T: transverse plane, S: sagittal plane)

Since the proposed system showed a promising performance for gait analysis, it will be applied to actual patients to monitor and analyze their motions.

Acknowledgment

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References

[1] U.S. Census Bureau, "The Elderly Population," [Online]. Available: http://www.census.gov/
[2] Parkinson Disease Foundation, "Statistics for Parkinson's Disease," [Online]. Available: http://www.pdf.org/
[3] Stroke Center, "Stroke Statistics," [Online]. Available: http://www.strokecenter.org/
[4] D.J. Gladstone, C.J. Danells, and S.E. Black, "The Fugl-Meyer Assessment of Motor Recovery after Stroke: A Critical Review of Its Measurement Properties," *Neurorehabilitation and Neural Repair*, vol. 16, 2002, pp. 232–240.
[5] R.A. Keith, C.V. Granger, B.B. Hamilton, and F.S. Sherwin, "The functional independence measure: a new tool for rehabilitation," *Adv. Clin. Rehabil.*, vol. 1, 1987, pp. 6–18.
[6] C. Collin, D.T. Wade, S. Davies, and V. Horne, "The Barthel ADL Index: A reliability study," *Disability and Rehabilitation*, vol. 10, 1988, pp. 61–63.
[7] VICON, [Online]. Available: http://www.vicon.com/
[8] D. Farina, R. Merletti, M. Nazzaro, and I. Caruso, "Effect of joint angle on EMG variables in leg and thigh muscles," *IEEE Engineering in Medicine and Biology Magazine*, vol. 20, 2001, pp. 62–71.
[9] W. Lin and T.S. Buchanan, "Prediction of joint moments using a neural network model of muscle activations from EMG signals," *IEEE Trans. on Neural Systems and Rehabilitation Engineering*, vol. 10, 2002, pp. 30–37.
[10] P. Bonato, P. Boissy, C. D. Croce, and S.H. Roy, "Changes in the surface EMG signal and the biomechanics of motion during a repetitive lifting tas," *IEEE Trans. on Neural Systems and Rehabilitation Engineering*, vol. 10, 2002, pp. 38–47.
[11] J. Bae, K. Kong and M. Tomizuka, "Real-time Estimation of Lower Extremity Joint Torques in Normal Gait," *in Proceedings of IFAC Symposium on Robot Control (SYROCO)*, pp. 577-582, 2009.
[12] S.K. Agrawal, S.K. Banala, A. Fattah, V. Sangwan, V. Krishnamoorthy, J.P. Scholz, and Wei-Li Hsu, "Assessment of Motion of a Swing Leg and Gait Rehabilitation With a Gravity Balancing Exoskeleton," *IEEE Trans. on Neural Systems and Rehabilitation Engineering*, vol. 15, 2007, pp. 410–420.
[13] J. Bae, K. Kong, N. Byl and M. Tomizuka, "`A Mobile Gait Monitoring System for Abnormal Gait Diagnosis and Rehabilitation: A Pilot Study for Parkinson's Disease Patients," ASME Journal of Biomechanical Engineering, vol.133, 2011, pp. 041005 (11 pages).
[14] K. Kong, and M. Tomizuka, "Sensor-Embedded Shoes Based Gait Monitoring System for Rehabilitation," *IEEE/ASME Trans. Mechatronics.*, vol.14, 2009, pp. 358–370.
[15] 9 Degrees of Freedom-Sensor Stick, [Online]. Available: http://www.sparkfun.com/
[16] Arduino Pro Mini, [Online]. Available: http://www.sparkfun.com/
[17] XBee 1mW Chip Antenna, [Online]. Available: http://www.sparkfun.com /
[18] VICON, [Online]. Available: http://www.vicon.com/
[19] J. Walrand and S. Parekh, "Communication Networks: A Concise Introduction," *Morgan and Claypool Publishers*, 2010.
[20] N. Phuong, H. Kang, Y. Suh, and Y. Ro, "A DCM Based Orientation Estimation Algorithm with an Inertial Measurement Unit and a Magnetic Compass," *Journal of Universal Computer Science*, vol. 15, pp.859-876, 2009.
[21] J. Perry, "Gait Analysis," *SLACK Inc.*, 1992.