The Development of a Portable Motion Analysis System Using FES Motion Sensors.

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Introduction

The use of accelerometers to measure human motion has been suggested for over 30 years (Smidt et al, 1971). With recent advances in miniaturization and the integration of rate gyroscopes, portable motion sensors are becoming a viable alternative to optical motion analysis systems (Mayagoitia et al, 2002). The development and refinement of accelerometer based motion sensors were largely driven by the needs of FES designers who needed real-time information about limb position for precise FES control. It was important that such sensors be ultra light, unobtrusive, and reliable (Williamson & Andrews, 2000; Simcox, et al, 2001).

Neopraxis Ltd developed sensor packs that are part of the Neopraxis FES-22B stimulator system (Simcox et al, 2001). Individual sensors are extremely compact (12 x 34 x 65 mm), and light (25 g) (Simcox, et al, 2001). The sensors collect segmental orientation information (tilt, angular velocity and accelerations) in two orthogonal planes. Data are collected using a Pocket PC (Casio Cassiopeia EG800) fitted with a compact flash serial interface card. Multiple sensors can be attached to various limb segments and connected in series with the Pocket PC. This configuration allows for unprecedented portability and flexibility. The sensors also allow for attachment of external analogue devices (i.e. foots switches).

The purpose of this study was to ascertain if the sensors, which were designed for FES, can be adapted as a multipurpose portable motion analysis system. Specifically, we report work in developing and testing the sensors for gait analysis and balance assessment.

Methods

Gait Analysis: The methodology used to validate the sensors against a 3D camera-based motion analysis system (CMAS) has been reported by Simcox et al (2001). However, while Simcox et al, (2001) used 5 sensors, we used 7 and two pairs of foot switches (Force Sensitive Resistors, MIE Ltd, Leeds, UK) to indicate initial contact (IC) and end-contact (EC) events. IC or EC events allow sensor data to be synchronized with the CMAS (Motus, Peak Performance, Englewood, Colorado, USA). The Motus CMAS uses 6 infrared sensitive cameras with a sampling frequency of 50 Hz. Sensor system sample rate was 27 Hz. We tested 7 able bodied subjects. Reflective joint markers were attached bilaterally on the following landmarks: 2nd metatarsal, lateral malleolus, tibial tuberosity, greater trochanter and glenohumeral joints. Segmental distances were measured between bony landmarks. Individual sensors were attached bilaterally on the ventral surfaces of foot, shank and thigh segments midway between the joint markers. One sensor was attached on the dorsal surface of the spinal column (L2-3). Two individual foot switches were attached to the plantar surface near the big toe and at the heel of each foot. The foot switches were connected to the left and right foot sensor. The toe and heel switch voltages were monitored individually so that EC and IC events for each foot could be sampled individually. All seven sensors were connected serially to a Pocket PC (Cassiopeia EG-800, Windows CE). Trial data were stored by the Pocket-PC as log files. At the end of testing session, individual log files were downloaded onto a PC for processing and analysis.

Protocol: At the beginning of testing, subjects were asked to stand as upright as possible with their back and heels touching the wall. Data were collected for 10 seconds. This was necessary to estimate sensor offset as the sensors, which are attached on the surface of each segment, are not parallel to the virtual segmental lines joining the proximal and distal landmarks. To obtain an estimate of the offset we assume that when a subject stands against the wall the thigh, shank and torso segments are orientated close to vertical (90°). The ankle joint is positioned neutral (0°). The offset readings are averaged and are subtracted in the gait analysis program. We acknowledge that our method to correct the offset may not be the most precise but we feel it is important to develop an easy and quick gait analysis system where an uncomplicated setup is an advantage.

Subjects were asked to walk at a comfortable walking pace along a 10-meter walkway. The calibrated space within the camera field of view is approximately 4 meters long, 1.5 meters wide and 2 meters high, enough for 2 to 3 strides. Between three to 10 walking trials were collected.

Analysis: To analyze the gait data we developed special software in Labview (6i). We report data from the sagittal plane. Segmental tilt angles are converted into real world coordinates based on the individual's anthropometric data and reconstructed into a biomechanical stick figure. Our program is based on frame-by-frame analysis and, combined with foot switch data, can analyze joint motion (ankle, knee, hip angles, angular velocity etc) and determine common spatiotemporal parameters (gait velocity, support phase, swing phase etc). Data from 3 right and 2 left strides were compared between CMAS and sensors. To compare data collected from different sampling frequencies both data sets were normalized to one gait cycle and spline interpolated to 100 points. The following data were analyzed: bilateral hip, knee and ankle joint angles, and gait velocity.

Balance Screening: The Neopraxis sensors are potentially ideal for the assessment of balance as they can directly measure sway. Clinical balance, or amount of sway, is assessed by measuring the center of pressure (COP) on the force plates. This approach assumes that the COP is representative of the total center of mass (COM) and represents sway. However, there are theoretical problems with this assumption. Sway by definition (Eng & Winter, 1993) is the movement of COM over the base of support and erroneous interpretations result if COP is used to measure sway (Eng & Winter, 1993). One sensor can measure sway directly if a sensor is placed on the dorsal aspect of torso close to the location of the COM (L2-3). The sensor measures angular deviations of the torso/COM in the sagittal and coronal planes. In this study, we measured sway by placing one sensor close to the location of the center of mass on the dorsal aspect of the torso (L2-3). Subject's sway was assessed using a well-known clinical sensory organization test, called modified CTSIB.

Protocol: Ten normal subjects were tested under four conditions: standing on a firm surface with eyes open and then closed, standing on 5-inch thick foam with eyes open and then closed. One sensor was firmly attached on the dorsal surface over the vertebral column of the subject's torso between levels L2-3. A small analog thumb trigger was connected to one connector port of the sensor. The subject was asked to stand still with one hand against the wall to increase stability. The experimenter stood behind the subject and started data collection. Sampling frequency was 57 Hz (18ms). After 5 seconds, the subject was asked to release their hand from the wall and start one of four conditions. Depending on the condition, specific instructions were given i.e. "step on the foam and when ready close your eyes and stand as still as possible until instructed to do otherwise." Once the subject was ready, the trigger was depressed and 40 seconds of sway data were collected. The total test with setup lasted 3 to 5 minutes.

Analysis: Data were downloaded onto a PC and analyzed using a custom balance assessment program. The first 5 seconds of data (hand against wall) prior to each balance condition were used to calculate sagittal and coronal neutral angles. Sway data angles were subtracted from the neutral angles so that all deviations were relative to 0º in coronal and sagittal planes. Thirty seconds of sway data were analyzed after the trigger analog signal. The coronal signal was filtered using a low-pass single order Butterworth filter at 1 Hz. We found it not necessary to filter the sagittal signal.

The following indices of sway were calculated from sagittal and coronal tilt angles: root mean error (RMS), average angular velocity (d/s) and the sum of all absolute angular deviations over 30s. Each measure is an indication of amount of sway. Larger numbers indicate more sway.

Results

Gait Analysis: Figure 1 shows the average joint angles and standard deviations for 5 strides (3 right and 2 left) obtained from the sensors and CMAS of 7 subjects. There is excellent agreement in terms of the shape of the waveforms and sub events. In fact, the sensor consistently picked up features in the gait cycle that were not picked up with the CMAS (i.e 1st rocker of the ankle joint). However, there still is considerable offset in the actual angular values. Some of the offset can be attributed to virtual segmental lines (joining the joint markers) not being entirely aligned and the choice use of the marker model. We will probably need to develop a more precise offset calculation protocol. However, the sensor data appear to agree more closely with standard gait reference data. The cross-correlations range between .92 and .99 for all subjects.

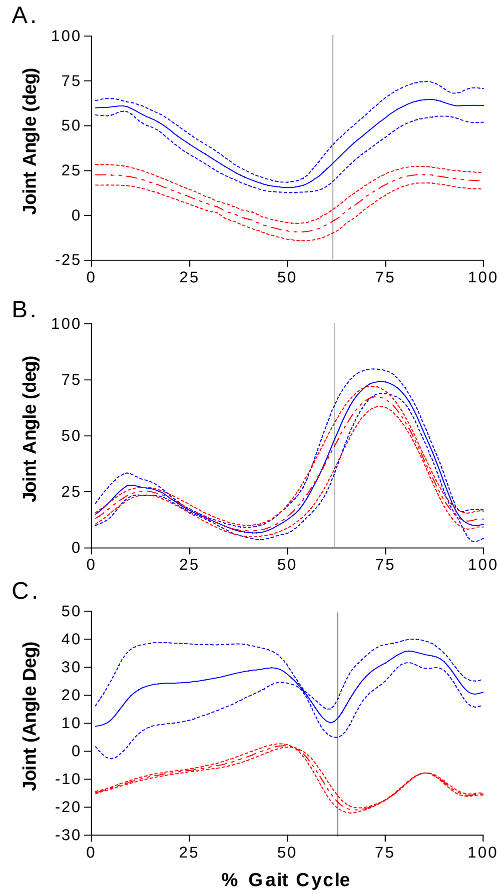

Figure 1. Ensemble average joint angles (3 strides left and right limbs of 7 subjects) graphs obtained from camera based (hashed lined) and the sensor system (solid). With standard deviation lines above and below. A= hip angle, B= knee angle, C= ankle angle. The vertical line indicates average toe off.

Gait velocity agreements were found to be excellent (1.45 ± .3 m/s sensors vs. 1.39 ± .2m/s CMAS), although the agreement depends very much on the accuracy of the anthropometric measures and marker placements. Event detection for CMAS was determined visually.

Balance Assessment: We report data for two sway scores: RMS and sway velocity (deg/s). Two 2-factor analysis of variance (2 Eyes Open/Closed x 2 Support surface) show that sway significantly increased as measured by RMS and sway velocity increased when subjects stood on foam (RMS: $F(1, 36)=23.842$, $p<.025$; Sway Velocity: $F(1, 36) = 22.082$, $p<.025$) as depicted in figure 2. There were no other significant differences. There was no detectable sensor drift. The results are consistent with what is reported in the literature. It suggests that the sensors are sensitive enough to detect sway of small magnitudes. The sensors are capable of resolving changes in tilt of .1º in both planes, which compares favorably with CMAS calculated COM. This demonstrates that the sensors are well suited to assess balance in a variety of assessment tests (i.e. tandem walk, one-legged stance, limits of stability). In addition, balance data can be collected quickly and at a fraction of the cost of current force plate systems. In short, it is a perfect clinical tool.

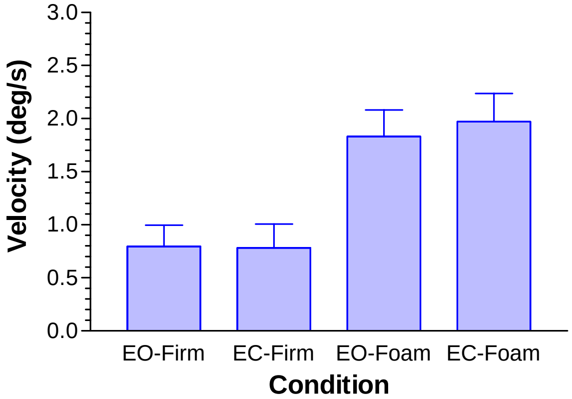

Figure 2. Average sway velocity score of ten subjects for the 4 sensory conditions obtained using one sensor attached to the torso (30s trial). Only the effect of support surface was statistically different.

Discussion / Conclusion:

The results show that the sensors are well suited as a versatile and cost-effective multipurpose motion analysis system; multiple sensors can be used for gait analysis, while a single sensor can be used to assess balance. For each application, the sensors report valid and reliable data. The sensors do not require extensive laboratory facilities and can be used almost anywhere. At the same time, we have also identified some limitations, for example, rigorous activities such as running are beyond the current capability of the sensors system hardware.

However, the sensors form a solid design concept for a completely portable and integrated motion analyzer. Specifically, the sensors can also be integrated with other devices, such as heart rate monitors and foot pressure distribution systems, allowing kinematic data to be correlated with other measures (kinetic and physiological). Obviously, such a system, which does not need dedicated laboratory facilities, will have many potential applications in clinical, rehabilitation, sporting and research domains.

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