CMAS: Clinical Movement Assessment System for Neuromotor Disorders

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Abstract

We are developing a clinical movement assessment system (CMAS) to quantify motor effort following neuromotor injury. The system will allow for easy and versatile data collection by non-technical staff and patients from a variety of devices by using recent advances in technology for real-time reconfiguration. We will demonstrate the effectiveness of our system by developing a patient-friendly device. A prototype of the device has been implemented and used to measure finger forces in flexion and extension. The efficiency of the CMAS devices will be demonstrated by making quantitative assessments of neuromotor efforts following stroke in a patient population. At present, our experimental results on two subjects show the effectiveness of our device.

1. Introduction

Emerging technologies are generating opportunities for improving medicine and public health. We are currently focusing on using recent advances in wireless networks and real-time reconfiguration to improve diagnosis and treatment of neuromotor impairments.

Strokes, spinal cord injuries, traumatic orthopedic injuries, and other disorders commonly lead to a loss of motor functions of the upper limbs. Currently, clinical assessments of motor functions are routinely scored subjectively or using ordinal scales [1-3]. New treatment and rehabilitation strategies are creating a need for more accurate and quantitative measures of motor function that can be made readily and accurately [4, 5]. Such Standardized, quantitative functional measurements would contribute to understanding neuromotor injuries. Moreover, quantitative measurements of motor function are the first step in developing appropriate population databases to enable rapid, patient-specific assessments of motor function.

The architecture we propose is a mobile and low profile system that may be used in home settings and can potentially assist stroke survivors with several therapy exercises.

2. Related work

Many studies have characterized individual finger function and coordination in both unimpaired and impaired populations [6-12]. Quantitative measurements of maximum force, time course of force onset and offset, and finger coordination yield valuable information about neuromotor function following injury, and could be very helpful in describing the characteristics of use-dependent plasticity during recovery. Although most current efforts are primarily limited to a research setting, there is growing interest in using quantitative measurements of function as clinical tools [13-17].

3. Hardware design

Our CMAS is based on a system architecture called CustoMed (Customizable Medical Monitoring device) developed at UCLA [18] . CustoMed is a platform for health monitoring using wireless sensor networks (WSN). This architecture is a network enabled system that supports various wearable sensors and contains on-board general computing capabilities for executing individually tailored event detection, alerts, and network communication with various medical informatics services. Our platform is composed of sensors, "med nodes" and a pocket PC. The most important component of our system, the med node, is a stand-alone component consisting of a processing unit and a battery which can support various types of sensors for physiological reading from human body. The customization of such system with a large number of "med nodes" is extremely fast even by non-engineering staff. In addition to support of various wearable sensors, the software downloaded onto the devices can be customizable over-the-air. Depending on sex, age, medical condition, and other variables, the software downloaded onto the devices differs. The idea of customization has not been emphasized before, but is an important concern, if the system is made to be robust enough to handle many different needs, as well as unexpected needs that may arise.

3.1. Sensors

CustoMed can interface with various types of sensors, such as pressure, galvanic skin response, flex, piezo-electric film and temperature sensors. These sensors can provide continual physiological measurements as well as environmental measurements, even when people operating in hazardous environments. In our first CMAS prototype, we utilize pressure sensors to measure the force applied by individual hand digits to a handgrip. Pressure sensors are ideal for measuring forces without disturbing the dynamics of a test. They can be used to measure both static and dynamic forces. They are thin enough to enable non-intrusive measurement. The resistance of the sensor decreases as force is applied.

3.2. Handgrip device

The CMAS handgrip device is capable of measuring forces generated by individual fingers using pressure sensors. The resistance against squeeze force in our handgrip device is adjustable. A picture of the handgrip device along with sensors and a med-node is shown in Figure 1.

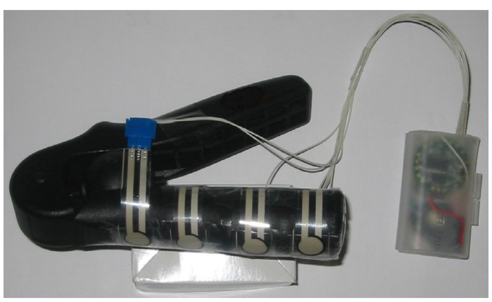

Figure 1. Handgrip device

3.3. Med node

Med Nodes enable the system to be flexible. Although their basic structure can remain fixed most of the time, customization of the system with a large number of "med nodes" is extremely fast. The "med nodes" can be complex enough to suit a range of applications, such as tracking knee motion after knee surgery to aiding Alzheimer patients residing in assisted living homes by detecting arousal and/or agitation (by measuring skin conductance). Furthermore, they support a variety of analog and digital sensors. The block is software programmable and can be customized for various applications and sensors. On-chip memory blocks are also available for data storage. The processors of "med nodes" are dot-motes developed at the University of California, Berkeley. A "med node" along with a handgrip and pressure sensors is shown in Figure 1.

3.4. Pocket PC

A pocket PC is responsible for collecting data from "med nodes" and classifying them. It dispatches the critical events detected by "med nodes" or the pocket PC, itself, to the Internet. Moreover, it coordinates and controls the overall functionality of the system. It can as well perform resource management to accommodate several objectives such as optimizing the power or enhancing the fault-tolerance. In many medical monitoring applications, pocket PCs can facilitate interactions with patients with their large displays while the mobility is not sacrificed.

4. Software design

4.1. Med node

The software for med nodes has been developed in C and under SOS, a new operating system for mote-class sensor nodes that takes a more dynamic point on the design spectrum [19]. SOS consists of dynamically-loaded modules and a common kernel, which implements messaging, dynamic memory, and module loading and unloading, among other services. Dynamic reconfigurability is one of our primary objectives. In the domain of embedded computing, reconfigurability is the ability to modify the software on individual nodes of a network after the system has been deployed and initialized. This provides the ability to incrementally update and add new software modules to med nodes after deployment, and remove unused software modules when they are no longer needed. This in particular is beneficial when the medical staff ought to remotely update the software that runs on med nodes. The ability to reconfigure the software on med nodes may potentially enhance the power consumption of the system (and the lifetime) by limiting the wireless transmission to only the most valuable data. In tiny embedded systems that we utilize -- Berkeley dot-motes-- the most power hungry component is the radio module. In addition, the local processing in med-nodes may be dynamically altered based on individual's need.

4.2. Pocket PC

The software on Pocket PC provides both serial connection with a gateway med node and Internet connection with an external server. The existing radio communication protocol on med-nodes is incompatible with the wireless receiver on Pocket PCs. Therefore, we utilize a base-station med node that collects information from all med-nodes and sends to the pocket PC through the serial connection. The base station med-node also facilitates the software reconfiguration (in SOS the new software modules that is installed must be dispatched from a base station). The Internet connection with an external server may be utilized for remote monitoring. As shown in Figure 2, in the first window of CMAS terminal on pocket PC, the user specifies the proper port configuration for both serial and Internet connections. When the external server is not available, the user may choose to bypass the Internet connectivity.

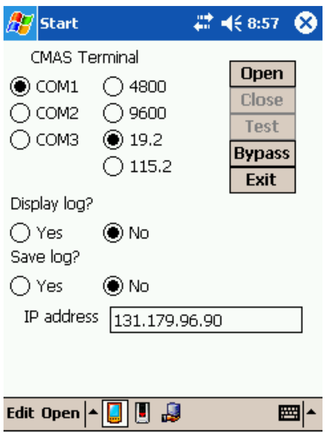

Figure 2. Connection setup window

For the particular handgrip device we utilize, four pressure sensors are employed. Hence, it is imperative that patient's hand digits are placed properly on individual sensors. In addition, due to the variable physical capabilities of patients, the applied force must be normalized. Hence, the screen shot shown in Figure 3 is designed to assess the maximum voluntary contraction (MVC) and ensure that the fingers are properly placed on the sensors.

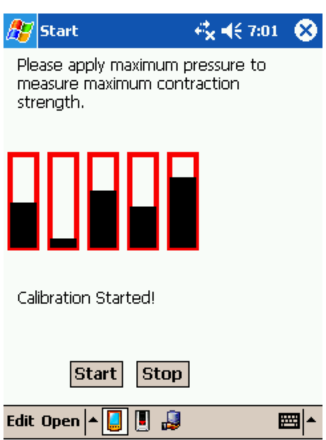

Figure 3. Calibration window

Following the calibration process, the main window of CMAS terminal is displayed as shown in Figure 4. The top window features a guide or a target cursor that moves on a sinusoidal wave (Fig. 4, black dot). The patient's response (i.e. the average forces applied to the pressure sensors) is displayed in the same window (Fig. 4, blue dot bounded by horizontal lines). The objective is to have the patient follow the guide by applying appropriate force to the handgrip device. The force may be scaled to 80%, 60%, 40% or 20% of the MVC force. Moreover, forces applied by individual fingers are illustrated on the CMAS terminal in four separate windows. Finally, the patient can choose from several test sets which will be described in the next section.

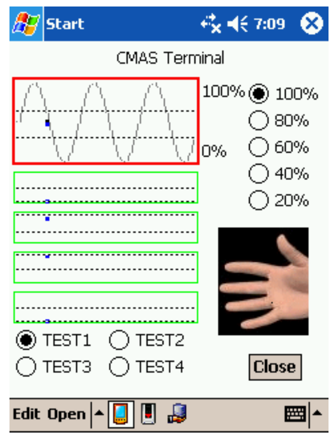

Figure 4. CMAS task window

5. Test sets

Four test sets have been developed for the CMAS:

- Test 1: The guide or the target travels on a sinusoidal waveform between zero and 100% of the patient's maximum strength. The period of the sine wave six seconds.
- Test 2: The objective in test 2 is to evaluate the fatigability in patients. The target moves between 20% and 40% of the MVC with the period of 200ms. The patient is expected to follow the indicator by rapidly squeezing the handgrip device.
- Test 3: This test is developed for evaluating the ability to finely modulate force production. The guide moves between 15% and 30% of the MVC on a sinusoidal wave with period of six seconds.
- Test 4: This test is the high strength fine grain motor control. This test is similar to test 3 except that the guide moves between 45% and 65% of the MVC.

5. Experimental results

Preliminary experiments conducted on unimpaired subjects (N=2) showed that subjects were able to finely modulate finger forces to achieve a desired average force (Figure 5). The variance in errors between average force and target forces were 6.4 ± 1.0%, 0.9 ± 5%, 38 ± 0.3% and 0.3 ± 0.3% of the guide variance for Task 1, 2, 3 and 4, respectively for both subjects over 6 trials. Subjects precisely matched the guide frequency (frequency differences < 0.005 Hz for all trials). Crosscorrelating the guide signal with the average force signal revealed that average forces from subjects lagged the guide forces by 80 ± 100 ms and 300 ± 0 ms for tasks 2 and 3. For tasks 1 and 4, the time mean time lags of -50 ms and -21 ms fell well within the variance 170 and 470 ms, respectively.

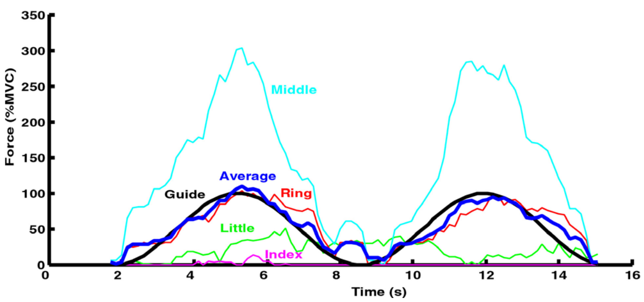

Figure 5. Average and individual forces during Task 1

Whereas subjects were able to precisely track the guide signals, they did not accomplish this by generating comparable forces with each finger (Figure 5). Large forces produced by middle fingers compensated for small forces generated by index and little fingers, reflecting an effective force synergy that maintained target average forces. The variance in errors between each individual force and the target forces is shown for every task and subject in Table 1. In addition, in frequency domain, the difference of the peak in the power spectrum between each individual force and the target forces is shown. Please note that each data point in Table 1 is the average taken over 3 trials.

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Task	Subject	Variance in errors with respect to the target force (%)					Frequency difference (Hz)				
		Average	Little finger	Ring finger	Middle finger	Index finger	Average	Little finger	Ring finger	Middle finger	Index finger
1	1	5.56	88.18	46.00	23.51	90.80	0.002	-0.085	0.000	0.002	0.001
1	2	7.34	77.40	30.58	57.48	99.26	0.002	0.003	0.002	0.003	-0.052
2	1	0.78	89.40	10.82	66.63	99.78	0.000	-0.023	-0.008	-0.002	-0.044
2	2	1.06	72.51	21.22	67.61	100.00	0.000	-0.025	0.000	-0.011	-0.049
3	1	61.15	93.55	44.39	54.84	98.34	0.360	-0.368	0.356	0.361	-0.037
3	2	15.89	78.44	33.18	55.23	100.00	0.003	-0.204	0.003	0.002	-0.683
4	1	0.44	88.74	32.95	40.56	86.11	-0.001	-0.022	-0.031	-0.022	-0.029
4	2	0.15	77.29	29.64	40.44	98.48	0.001	-0.035	-0.015	-0.010	-0.023

Table 1. Itemized force error and frequency difference