Research Article

Smart Wearable Systems for Enhanced Monitoring and Mobility

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ABSTRACT

The percentage of people over age 65 will shift from 12% to 20% nationwide while the average life expectancy for men and women of all races continues to rise, introducing a national and global concern for health related expenses. In particular, diminished stability leading to an increased risk of falling is on the forefront of medical expense projections. The World Health Organization (WHO) estimates there are 285 million suffering from visual impairment (39 million blind, 246 million low vision) worldwide. When adding the aging population with concomitant increases in life expectancy and the climbing rates of vision pathology, the numbers are even more dramatic. Blindness and low vision result in a host of social, emotional and health problems, often due to antecedent difficulties with mobility. This paper presents two smart wearable systems designed to enhance the mobility and monitoring of elderly and those with impaired vision. By using advances in sensors, actuators, and micro-electronics, these wearable systems acquire large amount of data, and with high speed data processing and pattern recognition, provide feedback signals to those wearing them. These systems are self-contained and operate with an easily accessible battery power. Details of the design and analysis of these smart wearable systems are presented.

INTRODUCTION

There are two major health related issues that prevent individuals from being fully mobile. One is falls and fear of falling, especially in elderly, and the other is vision impairment. In this section, some back ground information is provided that further describes fundamental health issues related to the causes of reduced mobility. The analysis and design of two smart wearables are presented in later sections to address these growing concerns.

By 2030 the demographics of the United States will dramatically change as the baby boomers retire. The percentage of people over age 65 will shift from 12% to 20% nationwide [Federal Interagency Forum on Aging Related Statistics, Key Indicators of Well-Being]. This coupled with a rising average life expectancy for men and women of all races has lead to a national concern for increasing health related expenses. In particular, diminished stability leading to an increased risk of falling is on the forefront of medical expense projections [Akyol 2007]. This is because approximately one-third of adults over 65 years of age report at least one fall per year [Okada et al 2001] and about 10% of these falls lead to serious injury, translating to $20 billion in fall-related medical expenses in 1994 and 2020 projections at or above $32 billion [Englander et al., 1996]. More troubling is that falling represents the seventh leading cause of death in persons older than 65 [Akyol 2007].

Several intrinsic and extrinsic factors are responsible for a loss in balance or fall. Frequently, it is an environmental obstacle that initiates the perturbation in balance (e.g. shift in center of mass), but the reduction in strength, cognition and sensory input of the aging may yield insufficient compensations (i.e., repositioning of center of pressure) and result in a fall(s) [Hahn and Chou, 2003]. It is the inherent decrease in functional reserves associated with aging that makes older people naturally more susceptible to falls, but other medical conditions can also contribute to the risk. Falls can be a sign of an acute illness [Fuller 2000], drug interactions, as well as metabolic disturbances, anemia, dehydration and cardiopulmonary disorders [Hill 2002]. A degradation in balance will correlate to a measurable functional deviation in gait, as shown in Figure 1.

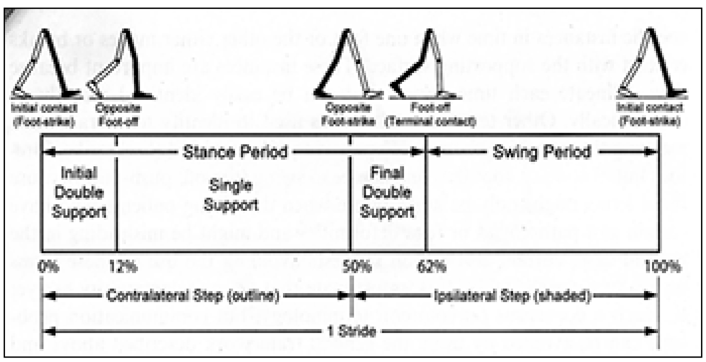


Figure 1: Representation of human gait cycle starting with initial heel contact of first foot then toe off of second foot. Swing phase of second foot followed by initial heel contact of second foot. Toe off of and swing phase of first foot and concluding the stride with heel contact of first foot [Carollo 2002].

For example, previous research has demonstrated that stride-to-stride variability, stance width, double limb support time and stride length are good indicators for reduced balance in community dwelling elderly [Maki 1997]. Therefore, we developed a smart shoe insole that monitors key gait characteristics and determines if the individual is in the danger of losing his/her balance and suffering from a fall.

The World Health Organization (WHO) estimates there are 285 million suffering from visual impairment (39 million blind, 246 million low vision) worldwide. When adding the aging population with concomitant increases in life expectancy and the climbing rates of vision pathology, the numbers are even more dramatic. Blindness and low vision result in a host of social, emotional and health problems, often due to the antecedent difficulties with mobility. Decreased mobility engenders reduced rates of employment and overall sedentary lives, generating concomitant medical problems, psychological illness and quality of life compromise; alarmingly, visual deficiency throughout the world significantly increases morbidity and mortality.

Reduced mobility caused by visual impairment restricts the ability of a large segment of the population from leading productive and fulfilling lives, in addition to increasing the incidence of injury and healthcare costs. Our long-term goal is to reverse this trend by providing wearable technology solutions to the visually impaired for improved functional independence. Our short-term product development goal in this project is to create a wearable device that allows a blindfolded end-user to safely navigate indoor spaces such as offices and homes, including navigating through and around doors, desks and people.

The current standardized mobility solution for the visually impaired is the 'white cane', a platform that was introduced in the 1920s. However, a cane is of limited functional use, given the low temporal and spatial resolution of the information it conveys. We have developed a device that remedies the cane's shortcomings, and in addition further augments the ability of visually impaired persons to both maintain balance and to localize objects in their environment. This device integrates a series of ranging and image sensors to extract pertinent information regarding obstacles and other objects in the environment, which will be displayed via a vibro-tactile code along the torso of an end-user.

Previous attempts at sensory augmentation in the visually impaired have failed for two important reasons: The first is technologic limitations resulting in devices that were heavy, cumbersome, had low processing capabilities, and that leveraged sensors with finite capabilities. The second is that previous devices did not convey environmental information in an 'intuitive' manner, consistent with the spatial processing capabilities of the central nervous system and characteristics of peripheral sensory neurons, rendering training periods lengthy and arduous. These problems are addressed in our novel platform.

WEARABLE DESIGNS

Two wearable systems are presented in this section. One is a Smart Shoe insole designed and developed for fall prevention. Another system is for mobility and navigation of those who are visually impaired.

Smart Shoe Insole:

This shoe insole with a robust early warning system can prevent elderly falls. Caregivers could gain faster feedback on drug interactions and therapeutic performance prior to a fall event or hospitalization. Fall intervention is especially important in nursing homes or assisted living environments where the rate of falls are approximately three times higher than the average for those in the community [Alexandr 2002]. To be feasible and practical, the smart platform technology would need to be cost effective enough to implement broadly within an assisted living community, sufficiently unobtrusive so as not to exacerbate existing balance problems, autonomous as so as not to add to a caregivers duties, and reliable enough to collect longitudinal gait data on a single individual. Gait data is frequently collected on known fallers, but not the general nursing home community. Far less frequently are variations in gait in an individual observed daily.

This smart insole integrates embedded microelectronics, sensors, signal processing, data extraction and signature development algorithms. This **Smart Shoe Insole** is a modular autonomous gait telemetry device, Figure 2. It is cost-effective shoe embedded technology that can continuously measure key features of gait, providing predictions on loss of balance and helping the user/caregiver with pertinent health-related information. The modular design is significant, as the same base platform can be integrated with temperature sensors to provide predictive information on foot ulcer formation in diabetics.

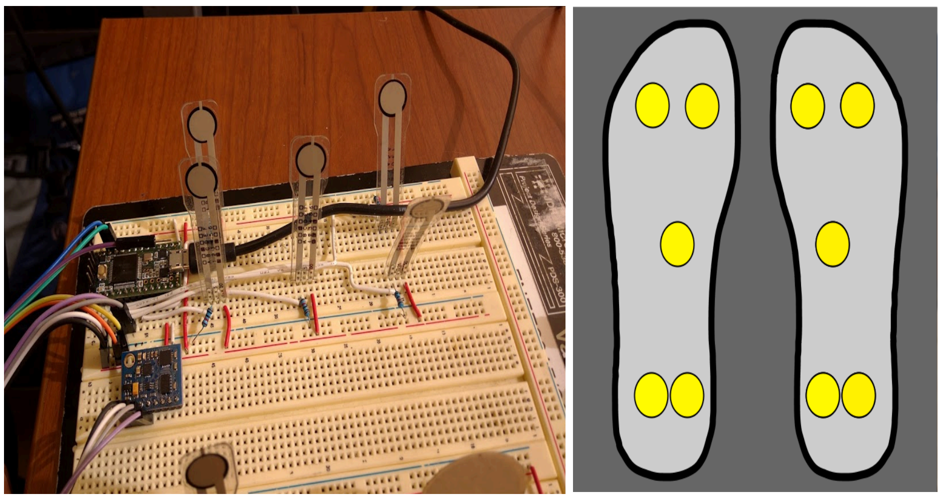


Figure 2: Smart Shoe Insole and Gait telemetry system

The sensor and communication system is embedded into the sole of a shoe with no external components (i.e. user can not perceive presence of this device). The embedded sensor system is controlled wirelessly by a base station dedicated to gathering, sorting and analyzing data collected from each shoe insole. The smart insole has multiple sensors used to measure different biomechanical signals, including pressure sensors as well as a 2-axis accelerometer used to measure acceleration in the medial-lateral and vertical directions. The smart insole consists of five main components. The first component is the sensor system that provides the human machine interface. Therefore, the accuracy of data is limited by the accuracy of our sensors. The measurement of the plantar pressure in a few regions of the foot has found to be a good indicator with regards to balance and gait [Carollo 2002]. These regions will include the heel and toe, which will be used to determine foot contact time and respective pressure distributions. The other regions will be inner and outer metatarsal heads which will determine stance width and their respective pressure distributions.

The next component is the electronic circuitry which is responsible for converting the output from the sensors into a measurable unit of voltage, filtering that voltage signal, and distributing power to the microcontroller, radio and sensors. It is important to realize that when the shoes are deployed to a specific individual their body weight can and will dramatically differ from others. Thus the circuitry also automatically and autonomously calibrates the gain on the individual sensors to provide and use the full voltage range available, thereby improving sensitivity.

The signal output from the circuitry is fed into the third component, the low power, micro-controller, which samples and stores the conditioned signals. The stored information can then be downloaded from the smart insole with the fourth component, a low power scientific and medical band radio which is also used by the computer to communicate with the smart insole. The radio has been set up so that it does not interfere with other devices, especially medical, operating in the ISM band.

To power the system, one lithium polymer battery was chosen. Lithium polymer batteries have an extremely high power to weight and power to volume ratio compared to wet lead-acid batters. Lithium polymer cells are also rechargeable, allowing for an excess of 500 charge cycles for the life of the battery providing a cost effective means of operation, and can provide high discharge. A protection circuit has been installed on each battery to prevent overheating and excessive current.

We have demonstrated measurement sensitivities suitable to detect modest (i.e., less than 10%) variations in an individual's walking characteristics. Based on this smart insole technology, we have developed an early warning system to monitor drug interactions, the onset of a physical ailment(s), and intervention success to circumvent falls, hospitalizations and other complications.

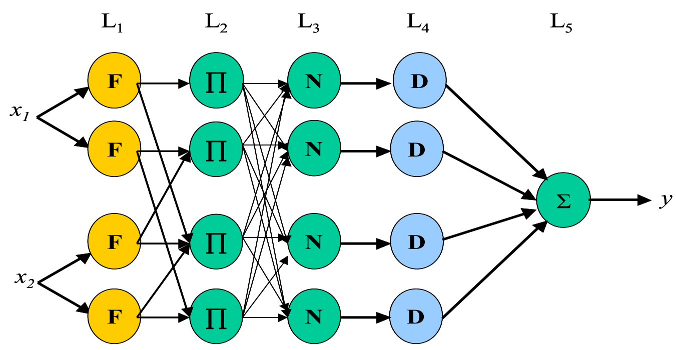
Fast feedback to the caregiver from a complex and large data set is critical to developing and implementing an intervention strategy in time to avoid a fall or other injurious event. This sets a requirement for the system's decision making support to operate reliably in a real-time environment. Based on our prior work [Shoureshi 2000, 2009], we have developed a neuro-fuzzy inference engine. The combination of artificial neural networks and fuzzy logic has demonstrated higher performance speeds, fast learning and adaptation, and possesses the human-like decision making process that is required to advert falls generated by gait instabilities and provide immunity for the proposed system in the presence of environmental variables.

We utilize mode shapes from spatial pressure data and two axis acceleration generated on the insole of a shoe to analyze static and dynamic stability using gain-weighting in conjunction with the rate of change of the pressure distribution shape and acceleration measurements (equation 1). In an electrical system gain-weighting can be designed into the electronics so that the output voltages already contain the weighting, but can also be done after measurements by simply adjusting the signal by a specified multiplication factor (equation 2).

Our previous research efforts [Shoureshi et al 2004] has resulted in a multilayer feedforward neural network, which uses Tsukamoto's fuzzy reasoning to generate membership functions in both fuzzification and defuzzification layers. This neuron-fuzzy network is used as the central processing element of the proposed structural nervous system. This neuro-fuzzy inference engine has five layers and can be used for any number of inputs and outputs (MIMO). It employs the gradient descent method and the least square estimation (LSE) algorithms to train the network. Figure 3 shows the architecture of this engine.

$$\Lambda(x,y)=\left(\langle x-x_2\rangle^0-\langle x-x_1\rangle^0\right)\left(\langle y-f_2(x)\rangle^0-\langle y-f_1(x)\rangle^0\right)$$
$$\begin{aligned}\text{where } x_1 &= \text{maximum positive x - location}\\ x_2 &= \text{maximum negative x - location}\\ f_1(x) &= \text{function describing surface in positive y - direction}\\ f_2(x) &= \text{function describing surface in negative y - direction}\\ \Lambda(x,y) &= \text{area of shaped sensor}\end{aligned} \tag{1}$$

$$\begin{aligned}S_i &= g_s\Lambda(x,y)\\ \text{where } g_s &= \text{gain function}\\ S_i &= \text{sensor shading factor}\end{aligned} \tag{2}$$



F : Fuzzification, **∏**: Multiplication,
N : Normalization, **D** : Defuzzification,
Σ : Summation

Figure 3: Structure of Neuro-Fuzzy Inference Engine

Wearable System for Mobility of the Visually Impaired:

Our platform integrates two core components, an outer wearable (a vest) and an inner wearable (a belt), in our current embodiment. The vest houses a complement of sensors providing three-dimensional obstacle detection mapping. The current embodiment leverages both SONAR (sound navigation and ranging) and RADAR (radio detection and ranging) technologies. We have multiple ultrasound sensors and one frequency modulated continuous wave (FMCW) RADAR sensor. These sensing modes, i.e., SONAR and RADAR, have overlapping mapping coordinates creating an important redundancy in our sensing strategy. As each mode has strengths in detecting objects in space, each mode also has introduced challenges. These limitations are best addressed by a multimodal approach that addresses inherent limitations, buttressing the inferiority of one sensor type with superiority of another sensor type in specific environmental constraints.

These sensors are embedded in customizable scaffolding that is retrofitted into a commercially available fleece vest. The scaffolding is composed of an HDPE (high-density polyethylene) thermoplastic material. The scaffolds are configured into a dual window design that can be adjusted rotationally in pitch and yaw for dynamic positioning on the body given the unique anthropomorphic differences of end-users.

Figure 4: Evolution of Wearable Platform Design. Upper left and lower left panels demonstrate a wearable platform with 8 ultrasonic sensors providing input to the tactile belt. The central and upper right panels demonstrate a wearable platform with 16 ultrasonic sensors and 1 RADAR unit and the corresponding tactile belt. The lower left panel demonstrates a visual display that 'reads out' the tactile output (for the benefit of the tester); in this scene, the unit is 'sensing' a solid object on the right hand side of substantial size, as indicated by the number of lights flashed, referring to the wall on the user's immediate right.

The sensors are mounted into the small window, which is mounted into the larger window, and placed into the vest. All of the wiring is strategically channeled through the two layers of the vest (outer shell and inner shell), along with the scaffolding, so only the 'head' of the sensor is exposed through small opening. Figure 4 shows several of the prototypes with different functional capabilities.

The belt houses a two-dimensional matrix of vibrotactile elements that 'buzz' based on environmentally relevant hazards. The belt is constructed out of a commercially available lumbar back support. The back support is made out of a high-density foam and neoprene material with female-male Velcro sections at each distal end. The vibrotactile elements are then sewn into the lumbar back support in a 1:1 relationship with the ultrasound sensors (the one exception is the central row of actuators responding to either the center right or center left column of sensors, another inherent redundancy). The vibrotactile elements are y-axis linear resonance actuators (LRAs) whose vibrations are directly perceived via the Pacinian corpuscle located in the hypodermal layer of the skin. The wiring is housed in a thin layer of artificial leather that is 2mm in thickness and positioned around LRAs in a simple 'mask.' The wiring is condensed for ruggedization to ribbon cabling. The belt is connected to the vest through an audio input/output channel that is positioned on the rear aspect of the belt once positioned on the body.

The sensor vest maps pertinent environmental obstacles in three-dimensional space, as shown in Figure 5, and the belt re-displays these spatially relevant hazards through the two-dimensional matrix of vibrotactile elements. This re-display is presented in a spatiotopically preserved manner, leveraging our innate body-centered frame of reference. In other words, vibratory warnings are displayed with preservation of relative horizontal and vertical positions, while the depth dimension is transformed into the frequency of vibration. The belt is positioned over the umbilicus as an anchoring point for conveying centrality. The central row of actuators is thus positioned directly over this body anchor, the extreme right row is positioned over the right flank in close proximity to the axillary line (armpit landmark), the extreme left row is positioned over the left flank in close proximity to the axillary line, and one intermediate actuator is placed approximately midway between the central anchor and the axillary line on each respective side to provide several columns of vibrotactile elements. Within each column, there are multiple actuators that are positioned three inches apart. This allows for many discrete points of contact with the body and can convey additional spatially pertinent information about verticality, in addition to more complex situational information.

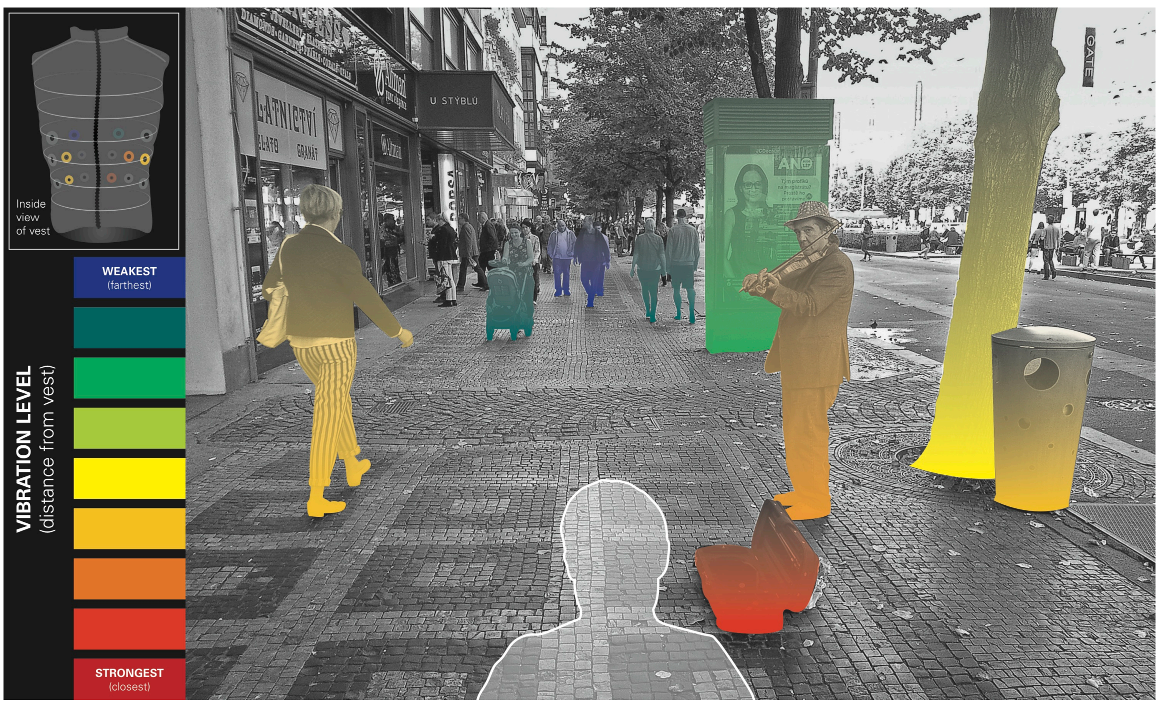


Figure 5: A scene deconstructed by our wearable system and converted into a depth map, translating the range of the object from the user to a specific frequency of vibration (near: more intense; far: less intense).

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