Health Monitoring Sensors for a Personal Mobility Aid for the Elderly

by

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Abstract

To accommodate the needs of seniors living in Assisted Living Facilities, MIT’s Field and Space Robotics Laboratory is developing PAMM, a walker-based robotic aid. Intended to facilitate mobility by providing support and guidance to the user, the walking aid is also to have health monitoring capabilities. This thesis describes the design of pulse and hydration sensors for integration on PAMM.

The development of a noninvasive, robust, ECG-based pulse sensor is presented, along with the design of the electronics and the algorithm used to process the signal. Experimental results demonstrate the overall competency of the sensor. Bioelectrical Impedance Analysis (BIA), a diagnostic tool that relates an individual’s hydration level to their body’s electrical impedance, is utilized to assess water imbalance. A new, qualitative technique is proposed to interpret the raw data obtained from BIA readings. An experimental set-up is designed and various validation tests are conducted to evaluate the new methodology. Considerations for future work are noted.
Acknowledgments

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Chapter 1

Introduction

1.1 Motivation

The average life expectancy in the United States has risen dramatically in the past century, from about 47 years in 1900 to about 76 years in 1996 (Department of Health and Human Services, 1997). This extension in lifespan results in a progressively older population. According to the U.S. Census Bureau, America’s population of seniors (age 65 or older) rose by 74% between 1970 and 1999. However, this growth rate pales in comparison to the surge that will take place over the next 30 years as America’s 76 million baby boomers age.

Various levels of residential care (home assistance, assisted living, nursing homes) have been established to accommodate the spectrum of needs required by the growing population of older people. With each progressive level of assistance, cost increases dramatically, with the maximum occurring at the transition from assisted living (at $25,000 per year) to a nursing home (at $90,000 per year) (Dubowsky, 2000). With elder care expected to become a 490 billion dollar industry by 2030, there are clear economic benefits to postponing a senior’s move from an assisted living facility to a nursing home (Greenwald, 1999).
1.2 The PAMM Project

An elderly person is transferred from an assisted living facility to a nursing home when their physical and cognitive abilities become so impaired that they require the continual attention of medical personal. It is proposed that a robotic aid be used to delay this transition. To demonstrate this idea, MIT’s Field and Space Robotics Laboratory is developing a series of PAMMs (Personal Aids for Mobility and Monitoring) to accommodate seniors living in assisted living facilities. The mobility aids are to provide guidance, support and health monitoring to the user.

The PAMM system concept is summarized in Figure 1.1. Mounted at the base of the handle bar, a force-torque sensor interprets the person’s directional intent. An onboard camera recognizes sign-posts fixed to the ceiling so the user’s whereabouts are continuously known. An array of sonar sensors provides the resident with assistance in crowded environments (Dubowsky, 2000).

![Figure 1.1: PAMM System Concept (Dubowsky, 2000)]
1.2.1 The SmartCane

Two test-beds are being developed to demonstrate the PAMM concept (see Figure 1.2). The first is a cane configuration with a skid-steer drive. Tested on a number of residents at a local assisted living facility it has received widespread user acceptance (Godding, 1999). The SmartCane’s success validates the legitimacy of the PAMM concept and the technologies developed to support it.

![SmartCane](image1.png)

![SmartCane Internal](image2.png)

Figure 1.2: PAMM Experimental Test-beds: SmartCane (left) and SmartWalker (right)

1.2.2 The SmartWalker

The SmartCane provides the user with limited mobility assistance. To offer greater physical stability to those with severe disabilities, a walker-based PAMM is being developed. Although the basic underlying technology of the cane is implemented into the
walker, two major features are added to this system. Skid-steer drives are replaced with omni-directional drives to give the user greater maneuverability in congested surroundings. Health monitoring sensors are also being implemented into the walker to provide information on the user's medical state.

1.2.3 Health Monitoring

Incorporating health sensors into the walker is advantageous because it allows for continuous monitoring. With health sensors mounted onto PAMM, medical information is recorded every time the user comes in contact with the walker. Since seniors with limited mobility capabilities cannot physically go anywhere without their walking aids, the sensors are never forgotten or left behind. Therefore, the walker collects data multiple times a day, every day. The resulting volume of data provides physicians with a clearer idea of the users' health. Using the person's medical history also simplifies the diagnostic process—only a change in a health signal is necessary to indicate the presence of an illness. Since the data is collected while the walker is in motion, the user's speed and applied forces are known. Thus, the person's activity level can be correlated with their health parameters, providing medical professions with further assessment capabilities.

1.3 Literature Review

In recent years, considerable research has been conducted to develop intelligent mobility aids for the disabled. The PAM-AID (shown in Figure 1.3) and the German RoTA are systems designed to provide mobility and guidance to the frail blind
The Nursebot and Care-O-bot have both been developed to assist the elderly living in private homes. The Nursebot concentrates on incorporating tele-presence, speech interface, and face tracking methods into its system (Baltus et al). While the Care-O-bot, still in the early stages of development intends to provide multimedia communication, operation of home electronics, and active support to its user (Schraft et al, 1998). Although these projects further the development of the technologies necessary for intelligent systems, none of the mentioned mobility aids offer on-board health sensors. The work presented in this thesis describes the development of health monitors for integration on PAMM.

1.4 Thesis Outline

This thesis is divided into five chapters. The first chapter introduces the
motivation and overall scope of the PAMM project. Chapter Two, describes the evaluative process used to assess user needs and selection of parameters to monitor. In the Chapter Three, the design and development of the heart rate sensor is discussed, along with the instrument’s hardware, software and experimental results. Chapter Four describes the fundamental research conducted to further the development of a hydration monitor. Results from validation tests are presented. Chapter Five summaries the thesis and makes suggestions for future work.
Chapter 2
Sensor Selection

2.1 Assessment of User Needs

To design a useful health monitoring system, it is important to understand the medical needs of the user. In this research, professionals who interact with the elderly on a daily basis (a rehabilitation physician, an assisted living facility director, a caretaker, a nurse, and a geriatric doctor) were consulted. Based on these interactions a list of possible parameters to monitor was derived (see Table 2.1). Some of the indicators, known as vital signs (blood pressure, heart rate and core body temperature) are considered to be standard means to gauge a person's health, while the others (glucose levels, hydration, gait, and pulse asymmetry) are parameters more relevant to older populations.
Parameters Suggested for Incorporation

<table>
<thead>
<tr>
<th>Parameters</th>
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<tbody>
<tr>
<td>Blood Pressure</td>
</tr>
<tr>
<td>Body Weight</td>
</tr>
<tr>
<td>Core Body Temperature</td>
</tr>
<tr>
<td>Fatigue Levels</td>
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<tr>
<td>Gait</td>
</tr>
<tr>
<td>Glucose Levels</td>
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<tr>
<td>Heart Rate</td>
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<tr>
<td>Hydration Levels</td>
</tr>
<tr>
<td>Pulse Oxymetry</td>
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<tr>
<td>Respiration Rate</td>
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</tbody>
</table>

Table 2.1: List of Health Parameters Suggested by Geriatric Specialists

2.2 System Specifications and Constraints

Incorporating a sensor into PAMM to monitor each of the suggested health predictors is an impractical task due to time, financial, and system constraints. Therefore, a set of an evaluative criteria is defined to narrow the list down. Each indicator is judged on medical significance and then on the instrumentation used to quantify it. The sensors must be inexpensive, maintain their functionality when mobile, and yield repeatable results. The instrumentation also must be noninvasive; wires and electrodes tend to frustrate and encumber the user. To exploit the structural and computational resources provided by PAMM, only on-board sensors are considered for implementation. Mounting sensors onto the walker, avoids the need for miniaturization (an issue plaguing many wearable sensors).
2.3 Parameters to Monitor

After preliminary assessment, five diagnostic quantities—body temperature, blood pressure, heart rate, hydration level, and gait—are selected. Upon further evaluation, however, only the latter three parameters are chosen for implementation.

2.3.1 Core Body Temperature

A vital sign of medicine, internal body temperature is one of the oldest indicators of a person’s well-being (Cromwell, 1980). The body’s efficient thermo-regulatory response maintains a constant (to within ±1° F) core body temperature by balancing heat generated by active tissues and heat lost to the environment (Guyton, 1976). If certain types of infection impair these control mechanisms, a fever can ensue. Skin temperature, a function of surface circulation and environmental temperature, can vary several degrees from point to point (85 to 95°F) (Cromwell, 1980). Therefore, non-contact devices, specifically infrared thermometers, are researched for incorporation with PAMM. This type of sensor measures temperature as a function of the radiation being emitted from a surface (Guyton, 1976). Although the technique shows promise, the cost of the instrument is found to be too expensive for integration into the walker.

2.3.2 Blood Pressure

Blood pressure is a physiological variable used to gauge cardiovascular function. It is characterized by two quantities: systolic pressure, defined when the heart contracts to pump blood, and diastolic pressure, measured when the heart is at rest in between beats (Moser, 1991). Elevated blood pressure, called hypertension, can lead to serious medical
conditions like arteriosclerosis, kidney damage, stroke and heart disease (National Institute of Health, 1994). Given the severity of the risks, a variety of ambulatory monitors are evaluated for implementation. The cuffs and accompanying wires used in these devices are found to be bulky and obtrusive to the user, and thus are not considered further.

2.3.3 Pulse

The average resting heart rate for mature adults varies from sixty to one hundred beats per minute (Guyton, 1976). When studied over time, a person’s pulse yields basic information about their health. An elevated pulse can suggest infection, fever, anxiety or high blood pressure, among other problems. During exercise, peak heart rates and recovery times give an indication of a person’s fitness level. Because of its assumed simplicity and widespread commercial availability, a pulse monitor is selected for incorporation with PAMM.

2.3.4 Hydration

Water maintains the homeostasis of the body’s internal environment for optimum cellular and tissue functioning. It provides a medium for transportation of metabolic products, a solvent for cellular reactions, and a means to dissolve nutrients. It also plays a critical role in the body’s thermoregulatory response (Dulbecco, 1997). Since fluid balance is needed for numerous physiological systems, an imbalance can often lead to an illness. Since geriatrics are especially susceptible to water loss, the user’s hydration level
becomes a desirable parameter to monitor. Due to a lack of noninvasive means to accurately diagnose fluid imbalance, this parameter is chosen for exploration.

2.3.5 Gait

A person’s gait, or pattern of walking, depends upon various physiological functions (Sudarsky, 1990). To walk, a person must maintain equilibrium and have adequate motor capabilities. Sensory systems, especially vision, are of increased significance when other modalities become impaired. Feedback from the senses yields information about the movement of the head and body relative to the surroundings (Craik and Oatis, 1995). Disease and the aging process can impair the performance of many physiological functions required for mobility. Therefore, gait monitored over time can be used as an indication of a person’s health. Characterized by stride length, stride frequency and velocity, gait can easily be monitored using data from the walker’s pre-existing encoders and force-torque sensor. No additional hardware is necessary. Preliminary testing, done by Shane MacNamara, demonstrates the ability of the walker to monitor the user’s gait (MacNamara, 2001).
Chapter 3
The Pulse Monitor

3.1 Chapter Overview

This chapter describes the design of a heart rate monitor for integration on PAMM. Although numerous compact and affordable pulse monitors are commercially available, these devices are found to be incompatible with the walker and so a robust sensor is created specifically for this application. Because of its insensitivity to motion disturbances, an ECG-based monitor is developed. Hardware and circuitry are designed to yield a clean pulse wave. A peak detection based algorithm is written to process the analog signal into a heart rate. The system is found to perform well in conditions similar to those in an assisted living facility.

3.2 Evaluation of Existing Devices

Microphone, infrared and piezoelectric based sensors are considered for implementation on the walker. Upon evaluation (detailed in Appendix A), the sensors are found to function reliably only in stationary environments. Mobility disturbances corrupted the readings. Since the pulse sensor must take readings while the walker is in motion, these devices are not considered further.
A final class of monitors, those based on detecting changes in surface potentials, is explored. For the heart to pump blood in an efficient manner, the atria and ventricles must contract in a coordinated manner. These contractions are indirectly initiated by action potentials, which raise the membrane’s potential from \(-80\) millivolts to \(+20\) millivolts (Guyton, 1976). As the sequence of these potentials transverse the heart, electrical currents spread into the surrounding tissues. A small proportion of these extend all the way to the surface of the body (Saladin, 1998). With the aid of electrodes, these electric potentials can be recorded and an electrocardiogram (ECG) can be constructed (see Figure 3.1).

![Figure 3.1 Normal Electrocardiogram (Guyton, 1976)](image)

Since most of the commercial ECG-based heart monitors require the user to wear electrodes, they are unsuitable for use with PAMM. However, heart monitoring systems found in exercise equipment are ideal for this application. With electrodes embedded in the handlebars of the device, the system is both noninvasive and robust. However, the pulse monitor circuitry and hardware used in exercise equipment cannot be separately purchased. So an ECG-based heart rate sensor is developed specifically for the walker.
3.3 Apparatus

The core of any ECG-based device is a circuit that filters and amplifies the signal, the first stage of which typically is a differential amplifier (Fleming and Feinberg, 1978). This is a crucial component since the bioelectric signal is one millivolt in peak-to-peak voltage, while power line noise, the major source of interference, can be orders of magnitude greater (Geddes, 1989). Figure 3.2 illustrates the use of the differential amplifier. The power line noise, coupled to the subject via the stray capacitances ($C_1$ and $C_2$), appears on the input terminals in phase, while the bioelectric signal appears on the input terminals out of phase with each other. Since the input terminals are of opposite signs (+ or -), only the difference in inputs will be amplified. With the line noise subtracted out, the output is a sole function of the bioelectric signal. Any common-mode input will be eliminated, not just power line noise. This ability greatly reduces the sensor’s sensitivity to mobility disturbances, a capability that the previous sensors lacked. The ability of an amplifier to reject common-mode inputs is characterized by its common-mode rejection ratio (CMRR), the ratio of response for a normal-mode signal to the response for a common-mode signal. In this circuit, an amplifier with a CMRR of 120dB is used.
After the signal passes through the differential amplifier, it is processed by two stages of filtering and amplification, the first of which is a first order high-pass filter. Although the component values yield a -3dB point of 15Hz, the high error tolerances in both the capacitors and resistor (10%-20%) contribute to a measured -3dB point of 0.65Hz. The high pass filter is included because it removes an observed low frequency component that distorts the signal. The output of this filter is then amplified 500 times and then processed by a first order low-pass filter. The measured cut-off frequency of this filter is 30Hz. This eliminates residual power line noise, along with any other high frequency (RF) contaminates. Finally, the signal is amplified 2.3 times and is sent through a capacitor to eliminate DC offsets. A circuit diagram of the electronics described in this section is presented in Figure B.1 in Appendix B.

An experiment is conducted to test the effectiveness of the circuit (see Figure 3.3). To detect the user’s voltage potential, two electrodes are inverted and placed under a square piece of aluminum foil. To simulate PAMM’s handle bars, the foil squares are
fixed to a wooden bar. The signals from each sensor are fed into the circuit. An oscilloscope is used to record the resulting waveform. Data is recorded while the subject stands still with her palms placed on the sensor pads. As one can note from Figure 3.3, the set-up yields a clean, useable signal in which the pulse wave can plainly be identified.

![Experimental Set-Up](image)

**Figure 3.3:** Experimental Set-Up (left) and Results (right)

### 3.4 Peak Detection Algorithm

The hardware yields an analog pulse wave. Software was written to process the signal into a heart rate. Fourier transforms were initially considered as a means to determine the pulse rate from the raw data. However, the sampling rate dictated by the computational limitations of PAMM (100Hz) yields power spectra too noisy to determine the user’s pulse. Therefore, a peak detection based algorithm was created to process the data. It calculates the user’s pulse every six seconds by identifying the voltage peaks,
averaging the time intervals between the peaks, and then converting that average value into a heart rate.

Written in *Matlab*, the code begins by filtering the raw data with a fourth order, low-pass Butterworth digital filter. With a cut-off frequency of 3Hz, the filter rids the signal of unwanted distortions due to noise in the transmission lines. The filtered data is then differentiated to remove DC offsets. Although this might seem to deteriorate the signal by increasing the occurrences of high frequency noise, this action actually emphasizes the peaks. As shown in Figure 3.4, differentiation tends to increase the uniformity of the noise, and so peaks are inadvertently accentuated.

![Figure 3.4: Processed Pulse Signal](image)

A peak occurs when the voltage potential crosses the threshold level. Since peak voltage levels vary from person to person, threshold values are not fixed. They are
determined by the standard deviation of the data multiplied by a constant (a constant of 2.45 is found to be optimum). Not only does the peak voltage vary between person to person, but as Figure 3.5 shows, it can also vary within the same individual. To account for this, thresholds are recalculated after every two seconds of data collection.

![Figure 3.5: Processed Pulse Signal with Thresholds](image)

Peak voltages are stored, while all other data points are set to zero. As Figure 3.6 shows, voltage maximums are now clearly distinguishable. Since the sampling rate may be faster than the time it takes for the primary peak to rise and fall, multiple data points may be recorded for a single peak. A logic gate is implemented into the code to eliminate these occurrences.
The time delay between each voltage peak is used to compute the user’s heart rate. Therefore, the performance of the pulse monitor is highly dependent upon precisely selecting the signal maxima. Erroneous peak detection can result from hardware limitations and insufficient threshold levels. When the threshold is too low, a secondary peak is counted (see Figure 3.7), and when the circuitry isn’t sensitive enough, a peak is missed (see Figure 3.8).
Figure 3.7: Errors in Peak Detection when a Secondary Peak is Counted

Figure 3.8: Errors in Peak Detection, the Insensitivity of the Hardware Misses a Peak

Chapter 3: The Pulse Monitor
Subroutines are incorporated into the algorithm to account for these errors. The time interval between each peak is found and is then compared to a reference value. If the time lapse is 25% less than the reference, it is assumed that a secondary peak was counted. To compensate for this, the smaller time interval is added to the successive one. Similarly, if the time gap is found to be 180% greater than the standard, it is assumed that a peak is missed and the time interval is cut in half. Clearly, the effectiveness of the described subroutines are highly dependent upon the reference value. Since a person’s heart rate is not constant, this standard value cannot be a fixed number. Instead, it is determined every six seconds by looking at the matrix of computed time intervals. Similar values (within 92% of each other) are defined as the reference. This is a reasonable scheme since the time that elapses from beat to beat in (healthy) individuals do not change substantially.

Since the added subroutines do not account for all possible sources of error one final clause is added to the algorithm. The standard deviation is taken of the matrix containing the time lapses between voltage peaks. Since this statistical tool gives an indication of how close (or far apart) data points are from each other, it is used to determine if there are any outliers in the data. Various filters were put in place for given standard deviation values. For example, if the computed standard deviation is found to be greater than 14, one or more of the data points is considered to be substantially different from the others. To eliminate the outlier, a new matrix is created that stores only the data points that are greater than 83% and less than 117% of the reference.

Using the time interval matrix, the heart rate of the user is calculated. First, the average time interval is found and then a simple calculation is done to convert the
quantity from time per beat to beats per minute. The measurement is then stored for future use.

3.5 Experimental Tests and Results

The overall performance of the pulse monitor is assessed by subjecting it to various testing conditions. To obtain a baseline, the health sensor is tested while operating in a stationary environment (without mobility disturbances). As the simplest case, this is used as an opportunity to optimize the algorithm parameters. Fifteen healthy young people, whose resting heart rate varied from 57 to 92 beats per minute, participated in the study. The original test bed is used to collect cardiac potentials for one hundred seconds on each volunteer. The data is then processes by the peak detection algorithm. The participants’ pulse rates are also manually computed by visually counting the peaks in the signal and computing a heart rate. The pulse monitor is found to perform considerably well, with the two methods in agreement 96% of the time. Since the resting heart rate of PAMM’s intended users is substantially lower (50-65 beats per minute) than the resting heart rate of average young adults, the experiment was repeated on two eighty year old volunteers. When the automated and manually tabulated heart rates are compared, they are found to agree 98.2% of the time.

More realistic conditions are incorporated into the next series of tests. Since the completed PAMM system is unavailable for testing, the user interface (the wooden bar from the experimental set-up) is fixed to the handlebars of a swivel office chair to simulate the walker (see Figure 3.9).
The floors in assisted living facilities are typically of vinyl tiling or of a fine carpet. Therefore, data is collected while operating the chair in both environments. The disturbances introduced into the system by the flooring are quantified by piezoelectric film fastened to the wooden bar. As one can note from Figure 3.10 and Figure 3.11, the pulse signals are unaffected by the added disturbances. This is an expected result since the hardware (the differential amplifier), rejects common-mode inputs.
Figure 3.10: Pulse Monitor Operating Over a Tile Floor

Figure 3.11: Pulse Monitor Operating Over Carpeting
Although the differential amplifier isolates the signal from mobility disturbances, its response is not always ideal for this application. The maneuvering of PAMM is governed by a force-torque sensor. Applying a differential force (or torque) at the handlebars initiates movement in the intended direction. This differential force, caused by the user shifting weight from one handle bar to another, corresponds to a change in contact area on each of the health sensor pads. Dissimilar areas of contact result in unequal signal amplitudes. Noise can only be subtracted out when the signals from each of the two input sensors are of comparable amplitudes; otherwise the signal is corrupted and is unusable (see Figure 3.12).

Figure 3.12: Distorted Waveform Due to Differential Signal Amplitudes
Some minor hardware redesign is done to compensate for this effect. Since compliance in the handlebars would help keep the contact areas constant, the metal sensors are placed on ½” thick insulation foam. As Figure 3.13 shows, although there is some increase in noise, the effect of applying differential forces to the handlebars is sufficiently reduced for this purpose.

![Image of handlebars and sensors with a graph showingProcessed Signal and Differential Force]

Figure 3.13: Improved Waveform Due to New Set-up

3.6 Implementation into PAMM

The software is modified for integration with the PAMM system. This included converting the peak detector algorithm into Visual C++, and adding some minor subroutines to it. Although the code is tested and found to operate well, the integrated pulse monitor could not be evaluated at an assisted living facility because the entire PAMM system is not ready for testing.

Even without field trials, a deficiency in the sensor is realized. There would be instances when continuous monitoring is inappropriate and lead to meaningless results.
These situations occur when the user is away from the walker or (since the heart rate sensor requires two inputs) when only one hand is in contact with PAMM’s handlebars. Therefore circuitry is designed to generate a binary flag to indicate when the time is suitable to record the user’s pulse.

Fundamentally, the circuit (shown in Figure B.2 in Appendix B) is a noise detector. Since a person becomes an antenna for power line noise when in contact with conductive materials, the amplitude of the noise should increase when the material is touched. This occurrence, shown in Figure 3.14, is observed when a hand is placed on PAMM’s metal health sensor pads. The change in amplitude is the basis for determining the proper time to compute the user’s pulse.

![Figure 3.14: Noise Amplitude when Hand is Placed on Metal Health Sensor](image)

A comparator is used to detect the change in signal amplitude. When the noise exceeds the voltage threshold set by the comparator, the component outputs five volts. A
threshold of 460mV with a hysteresis of 200mV is found to be optimum. The operation of the comparator can be observed from Figure 3.16; once the hand is placed on the sensor the noise level crosses the threshold and the comparator outputs five volts.

![Comparator Output](image)

**Figure 3.15:** The Use of a Comparator to Indicate User Input

However, since the line noise is sinusoidal, it only exceeds the threshold level half the time, and so the comparator output oscillates between zero and five volts. To achieve a binary output, a self-compensating diode detector is used to rectify the signal. As shown in Figure 3.16, the rectified signal has a ripple. This is a function of the capacitor value used in the diode detector circuit.
Although it isn’t essential, another comparator is added to eliminate the ripple from the rectified output. This time, a threshold of 1.2 volts with 100mV hysteresis is used. As shown in Figure 3.17, circuit yields a clean binary output. Since comparators have inverting inputs, the signal is now active low (the voltage is zero when the sensor pads are touched). This is desirable since active low is the standard in digital electronics.
The described electronics is built for each sensor pad (see Figure B2 in Appendix B). The outputs from each circuit are fed into a logic gate. An OR gate is used because its output will only be low only if both inputs are low (see Table 3.1). This ensures that the user’s heart rate is only recorded when both hands are in contact with PAMM’s handle bars. This concept is illustrated in Figure 3.18.

<table>
<thead>
<tr>
<th>Input from Right Hand</th>
<th>Input from Left Hand</th>
<th>Output</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>0</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>1</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
</tbody>
</table>

Table 3.1: Logic OR gate Truth Table
3.7 Summary & Future Work

This chapter described the development of a robust, pulse monitor for use with PAMM. Because of its immunity to mobility disturbances, an ECG-based monitor was developed. Circuitry was designed to yield a clean, identifiable pulse wave. To convert the analog signal into an actual heart rate, a peak detection based algorithm was written. Tested rigorously, the system was found to perform well in conditions similar to those in an assisted living facility. As a final measure, the software was implemented into PAMM so that the user’s pulse could be continuously monitored.
Once the PAMM system is functioning reliably, the integrated pulse monitor needs to be tested in an Assisted Living Facility with the proposed users. Field trials should be conducted to validate the instrumentation. Additional research can be done to investigate how an individual’s heart rate responds to their level of activity. The pulse data could also be used as a feedback signal to evaluate user acceptance and the overall effectiveness of the control system parameters.
Chapter 4

The Hydration Monitor

4.1 Chapter Overview

Next to oxygen, water is the most essential nutrient for normal physiological functioning (Dulbecco, 1997). It provides a fluid medium in which nutrients are dissolved, blood components are transported, and intracellular reactions take place. Vital for thermoregulation, it comprises a thermal barrier to prevent hypothermia and heat stroke (Sadler, 1999). With the dependency of numerous cellular and tissue functions on water, insufficient fluid volume can result in an illness. Since seniors are susceptible to fluid imbalance hydration level is a desirable parameter to monitor.

Bioelectrical impedance analysis (BIA) attempts to provide a noninvasive tool to measure hydration. The underlying concept of BIA is that the electrical impedance of the body to a weak current is inversely proportional to the amount of total water in the body. Due to inherently erroneous assumptions, the current BIA technique yields inaccurate results. In this chapter, fundamental research is conducted to further the BIA methodology. A new practice, that looks for a change in a person’s voltage potential to imply fluid imbalance is explored.
4.2 Water Balance in the Body

An individual's total body water (TBW) is distributed among the intracellular fluid (ICF) and extracellular fluid compartments (ECF) separated by a selectively permeable membrane. Fluid moves through the barrier by way of osmosis. This mechanism is the net diffusion of water from a side low solute concentration to a side of high solute concentration (Saladin, 1998). Since the movement of water in or out of a cell is a clear function of the relative concentrations of solutes, the most common types of solute, electrolytes, play a critical role in water balance (Guyton, 1976).

The body accomplishes fluid balance by conserving fluids when there is a loss and expelling them when there are in excess. Appreciable water losses result in a decrease in blood volume. The corresponding rise in osmolarity triggers water from the tissue fluid to enter the bloodstream. In turn, the osmolarity of the tissue fluid rises and water moves out of the cell to compensate. The change in osmolarity of all three fluid components is sensed by the hypothalamus, and causes the antidiuretic hormone (ADH) to be released from the posterior lobe of the pituitary gland. ADH, (along with other stimuli), act on the thirst center of the hypothalamus, resulting in the production of smaller volume and greater viscosity of saliva. Inducing a sticky feeling in the mouth, this action helps motivate one to drink. The antidiuretic hormone also works on the kidneys by facilitating the reabsorption of water into the body so it can be returned to the ECF. Water retention causes a decline in urine volume, increasing the ratio of sodium to water.

The body responds to the ingestion of excess fluid by increasing fluid output. High blood volume and low blood osmolarity act to inhibit the release of ADH. As a
result, the kidneys reabsorb a decreased amount of water so to reduce total body water. With the increase in urine output, the ratio of sodium to water decreases.

Blood volume and osmolarity clearly play a critical role in water balance. This relationship is reflected in the electrolyte concentrations found in urine. The ratio of sodium to water in urine is commonly used as an invasive means to confirm fluid disorders.

A variety of conditions result from fluid imbalance. Dehydration is one disorder that results when the body eliminates significantly more water then sodium. The ratio of water to electrolyte loss occurs along a spectrum and is indicative of the severity of the condition. Prevalent within certain populations, dehydration is a common cause of hospitalization and death in the geriatric community. Statistics from a 1991 US survey of Medicare recipients revealed that almost half of the Medicare beneficiaries hospitalized with dehydration as the primary diagnosis died within 1 year of admission (Dulbecco, 1997).

Many factors contribute to the predisposition of geriatrics to fluid imbalance. As one ages, their water reserve by weight drops from 60% to 45% (Reiff, 1987). A mature kidney losses it capability to regulate water balance (Rowe and Besdine, 1982). And the thirst mechanism along with the thermoregulatory response also diminishes with age.

Considering the many functions of water in the body, any loss of water is profound, however, there are physiological consequences specific to the elderly. One of the primary functions of water is to dilute water soluble medication. Since the therapeutic effect and toxicity of medicine is related to its concentration within body fluids, a loss of water can alter the effective dosage and lead to further medical
complications (Reiff, 1987). Incontinence, a common geriatric disorder, may contribute to dehydration. Uncontrollable bowel or urinary movements cause some to self-regulate their dietary inputs. Insufficient water intake may result in occurrences of diarrhea and further compound instances of incontinence. Dehydration in the elderly can also contribute to orthostatics. Caused by a reduction in systolic blood pressure (of at least 20 mmHg) when rising to the standing position, orthostatics leads to inadequate blood flow to the brain. The condition predisposes individuals to dizziness and has been shown to contribute to falls (Tinetti and Speechley, 1989).

4.3 Evaluation of Bioelectric Impedance Analysis

Due to its importance in health, along with a growing elderly population, one’s hydration level becomes a desirable parameter to monitor. Bioelectrical impedance analysis (BIA) is the only noninvasive, quantitative tool available that attempts to measure water imbalance.

The underlying concept of BIA is that the electrical impedance of the body to a weak current is inversely proportional to the amount of total water in the body (Olde Rikkert et al, 1997). The more hydrated the person is the less resistant they are to the flow of an electric current. The impedance of a person is modeled as an isotropic conductor (Kushner, 1992). A function of its geometric dimensions, impedance is written as:

\[
Z = \frac{\rho L}{A}
\]

where

- \( Z \) = impedance (ohm) \hspace{1cm} (4-1)
- \( \rho \) = specific resistivity (ohm-cm)
- \( L \) = conductor length (cm)
- \( A \) = cross-sectional area (cm²)
Multiplying the left side of equation 4-1 by L/L:

\[ Z = \rho \frac{L^2}{AL} \]  \hspace{1cm} (4-2)

With A*L being equal to volume (cm$^3$), the equation can be rewritten as:

\[ V = \rho \frac{L^2}{Z} \]  \hspace{1cm} (4-3)

Therefore, electrical volume is computed by measuring the length and impedance of the conductor. If, as in this case, the electrical volume of a biological compartment is of interest, the impedance will be complex. As shown in Figure 4.1, the plasma membrane is modeled as a resistor and capacitor in parallel. The cellular membrane maintains an ion concentration gradient between the intracellular and extracellular space. This potential difference creates a capacitive effect, while the fluid models a resistive one.

![Electrical Equivalent of Plasma Membrane](image)

Figure 4.1: Electrical Equivalent of Plasma Membrane (Liedtke, 1997)
The impedance measurement is obtained by introducing a small (50-800μAmps), 50kHz electrical signal into the body and then recording the resulting potential difference (see Figure 4.2). At 50kHz the capacitive and resistive contributions are said to be equal and so the person's impedance can be easily tabulated. A regression equation is used to convert the person's impedance value into a measurement of their total body water (TWB) in liters (Rikkert, 1997).

Figure 4.2: Method of Bioelectrical Impedance Analysis

The problem with BIA is in the interpretation of the raw data. The algorithm used to translate the impedance measurement into TBW is derived by noting a person's age, height, weight and hydration level (measured via the BIA technique). The data is then curve fit to a reference method, such as isotope dilution. This analysis yields a regression equation with the most sensitive parameter being the person's height, not their impedance. Additionally, this methodology makes the erroneous assumption that the relationship between body composition and electrical impedance is uniform within and between individuals (Roubenoff, 1997). However, body composition is not constant, it
changes as one ages. Since the regression equations are developed within specific populations, inaccuracies occur when the equation is applied to individuals outside that population.

Although the basic theory is sound, the inconsistencies in data analysis warrant taking a new approach towards BIA. Knowing a person’s exact water content is not of critical medical concern, but a significant change is. Therefore, utilizing the same measuring technique as before, a small alternating current would be introduced into the body. However, instead of using the resulting potential difference as a means to acquire an impedance measurement so that a numerical value for TBW can be attained, the voltage signal would instead be examined for a change in magnitude or phase from previous recordings. It is proposed that a change in potential would result from a change in hydration, and thus be an indication of one’s health (see Figure 4.3).

Figure 4.3: Proposed Methodology to Detect Water Imbalance

Since readings can be taken each time the user operates the walker, this new methodology is ideal for use with PAMM. The details of the hardware necessary to
implement a hydration monitor onto the walker are not considered; rather it was of primary concern to first demonstrate the validity of this new technique as a means to adequately predict one’s hydration status.

4.4 Hydration Instrumentation

To evaluate the new approach, an experimental device is created. It consists of three major components: circuitry (to provide the high frequency constant current source), electrodes (to detect and impart the signal), and an oscilloscope (to record the voltage potential).

A battery-operated circuit is designed to provide the small, high frequency current source. Although various op-amp based oscillators were tried, an integrated component is used to achieve the necessary frequency of 50kHz. The chip, along with supplementary circuitry (adapted from a 1996 design by David L. Jones) provides a clean, tunable sine wave. Various electronic components are added to achieve the desired peak to peak voltage output and to make the circuit battery operated. Additional hardware is built to form a Howland current source (Horowitz and Hill, 1989). The signal remains constant (to ± 3%) when the load remains below 3000 ohms. This is a sufficient range of operation considering total body impedance at 50kHz is reported to be from 500-1000 ohms (Reilly, 1998).

Four disposable ECG electrodes are used to sense and impose the electrical signal. A tetrapolar configuration is employed to eliminate the capacitive and highly resistive effect (70 kohms) of the outer skin layer (Reilly, 1998). Since this medium is substantially less conductive than the longitudinal conductivity of the muscle, its current...
density will be very small. To a good approximation therefore, the equipotential surfaces
due to the exciting current lie perpendicular to the skin surface (Aaron, 1997). Then
(assuming the measuring device draws negligible current), voltage measurements along
the skin surface are only a function of the body’s internal components (see Figure 4.4).

![Figure 4.4: Schematic of Equipotentials and Current Flow Lines in High-Resistivity Skin Layer.](image)

Operating at a sampling speed of 250Mhz, an oscilloscope is used to record the
voltage potential. The scope’s preprogrammed measurement functions are utilized to
note the waveform’s peak to peak voltage and RMS value.

4.5 Experimental Protocol and Safety

BIA readings are taken by placing two electrodes on the ventral side on each arm
of the participant. The detecting electrode edge is placed on the upper part of the
forearm, while the signal electrode is placed 5 ½” below, just under the wrist. Electrical
leads connect the electrodes to the appropriate machine, the detecting electrode to the oscilloscope and the signal electrode to the pulse generator (see Figure 4.5). The circuit introduces a current of 450μA at 50kHz into the body (for a few seconds) while the voltage potential is recorded. The entire procedure takes a few minutes.

Since the protocol requires an electrical current to be introduced into the subject the safety of the procedure is thoroughly investigated. The assessment is based on the magnitude of current that can be deliberately introduced into the subject without harm. Due to the large magnitude of cellular membrane capacitance, sensation and pain thresholds increase an order of magnitude as frequencies increase (Geddes and Baker, 1989). Three studies report at a frequency of 50kHz, the perception threshold to be approximately 40 milliamps, while pain and involuntary muscle contractions occur at
higher currents levels (Chatterjee, 1986). This sensation threshold is 120 times the current used in the instrument. Furthermore, the use of batteries greatly diminishes the user from risk of macroshock. Therefore, the high frequency currents proposed by this experimental device are found to present no hazard to the subject. Although, before validation studies could begin, official approval had to be granted by MIT's Committee on the Use of Humans as Experimental Subjects (COUHES). A proposal and an informed consent form were submitted and permission was given.

4.6 Experimental Testing & Results

The challenge of this work is in the validation of the proposed technique as a reliable means to monitor one's hydration status. The purpose of the experiments is to disturb the subject's water balance in a controlled and measurable manner and see if the hydration readings reflect the change. A valid assessment can only be made if this objective is satisfied. Various methods, similar to those used in clinical dehydration studies, are employed to impose water loss. However, without the medical or financial resources to conduct independent hydration measurements (like isotope dilution) a water imbalance cannot be certain. Without this assurance it is difficult to make a conclusive statement about the functionality of the instrumentation.

4.6.1 Fasting Experiments

The first phase of testing attempts to induce a water imbalance with fasting. To ensure an empty stomach, the healthy young volunteer abstained from food and drink for fourteen hours, after which a meal was consumed. To detect the effect of the prescribed
regiment, readings are taken in fifteen minute intervals an hour before and two hours after food and drink were ingested. The experiment is repeated three times on the same subject to ensure repeatability. One would predict constant or slightly increasing (due to rise in body resistance) voltage readings prior to consumption, and a decline (due to a drop in body resistance) after digestion (assumed to occur about ninety minutes later). Also, since the experiment is performed on the same person the voltage readings are expected to be of similar numerical value. However, these results are not attained (see Figure 4.6).

Two reasons are hypothesized to explain why the induced water imbalance was not detected by the instrumentation. Since the testing is performed on a healthy, young adult, it was thought that either fasting didn’t disturb the system or the participant’s
kidneys are compensating for the imbalance. Another consideration is that the experimental device is not sensitive enough to detect a change in fluid volume.

To evaluate the basic functionality of the device, the system is isolated from the noted (possible) sources of error by conducting experiments on a passive system—beef. If a known change in water content cannot be sensed in this simple system, then it cannot be detected in the complex, dynamic human body.

4.6.2 Meat Experiments

The purpose of this phase of testing is to impose a known water imbalance on the meat sample and then see if the instrument is sensitive enough to reflect the change. The experimental procedure developed to test this objective required a considerable amount of refinement. That process is detailed in Appendix D.

Significant water imbalance is accomplished by hydrating and then dehydrating a consistently thin, piece of top round cut meat. Changes in water volume are quantified by fluctuations in the piece's weight (a scale with ± 0.01g resolution is used). The meat is hydrated by submerging it in saline solution of osmolarity similar to that of cellular fluids. The sample is removed from the bath every fifteen minutes so hydration and weight measurements could be taken. Once a 2% increase in body weight is observed, the meat is dehydrated in a self-constructed dryer. The drying mechanism (described in Appendix D) is made similar in function to those used to preserve foods, the critical intention being to desiccate the food, not cook it. The sample is removed from the dryer every fifteen minutes so hydration and weight readings can be recorded. The experiment is concluded when a 12% decrease in body weight is noted. As shown in Figure 4.7 and
in Figure 4.8, clear trends, which agree with published reports emerge. Hydration leads to a decrease in resistance (and in voltage potential) while dehydration causes an increase in resistance (and in voltage potential). These results demonstrate the instrument’s sensitivity to changes in water balance. With the assurance of this capability, validation testing began with human volunteers.

![Graph showing results from hydration of meat sample](image)

Figure 4.7: Results from Hydration of Meat Sample
Before experiments can be performed on people, a means had to be devised to deliberately induce fluid loss in a safe and appreciable manner. Sports medicine journals were consulted to determine acceptable means to dehydrate subjects. Three mechanisms: exercise, saunas, and diuretics, were found to be fairly standard. All three are used in this research.

4.6.3 Athletes

Exercise is the first means used to dehydrate the subjects. However, before formal testing began, the unknown impact of sweat on the BIA readings is explored. Although a tetrapolar electrode configuration is used, it was still of concern that sweat, due to its high concentration of NaCl, would radically lower the skin’s resistance and invalidate the
measurements. A simple experiment proves this apprehension to be unfounded. A BIA reading is recorded with electrodes on each arm. Another measurement is taken after each arm is soaked (for an hour) in a saline solution of osmolarity similar to that of sweat. Total body resistance is computed for each case and is found to differ only by 2.60%. The experiment is repeated, only this time to see the effect of disproportionate amounts of sweat, a reading from a dry arm to a wet (saline soaked) arm is taken. When compared to data recorded on two dry arms, total body resistance is found to vary by 4.95%. These deviations are considered to be negligible since truly dehydrated subjects are reported to show a change in resistance of 20% (Rikkert, 1997).

<table>
<thead>
<tr>
<th>Measurement Method</th>
<th>Percent Change In Resistance</th>
</tr>
</thead>
<tbody>
<tr>
<td>Wet Arm to Wet Arm</td>
<td>2.60%</td>
</tr>
<tr>
<td>Wet Arm to Dry Arm</td>
<td>4.95%</td>
</tr>
</tbody>
</table>

Table 4.1: Effect of Sweat on BIA Readings

This result is additionally confirmed by measurements taken by Osypka (1963). Using low voltages and large copper-cylinder hand-to-hand contacts, total body impedance is computed over the frequency range of 0.3 to 100kHz. The results, plotted in Figure 4.9, show that above 5kHz, skin hydration becomes insignificant (Reilly, 1998).
Since sweat isn’t found to introduce any considerable errors, an exercise-based dehydration experiment is conducted. Modifying procedures found in sports medicine journals, the volunteer ran (indoors) until a 2% decrease in body weight is observed. During the run, the intake of fluids was prohibited. To replenish lost water and electrolytes, once the exercise is finished the subject drank a 32oz sports drink. BIA readings, taken periodically taken during the test are plotted in Figure 4.10. A 20% decrease in voltage potential (which is recovered after fluids are ingested) is found. This result is significant enough to warrant further, more complete experiments.
A more comprehensive hydration study is conducted with MIT's men's cross-country track team. Sixteen students, each running a minimum of fifty miles a week (for at least five months) agreed to partake in the experiment. Their participation was compensated with five dollars. As before, approval to conduct the test approval was granted from COUHES.

Before their outdoor scheduled run began, the volunteers' weight and voltage potential are measured. Before the readings are taken the lower arms of the participants (where the electrodes were placed) are cleaned. So water imbalance could be measured by changes in body weight, the intake of fluids is prohibited once the exercise began. After practice, the procedure is repeated. Once again, the subjects' weight and hydration readings are recorded. The results, shown in Figure 4.11, are completely chaotic. The percentage changes in potential are not even of the same sign. However, each of the

Chapter 4: The Hydration Monitor

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runners’ percentage weight loss is significantly under the guideline of 3%-5% used by the Academy of Pediatrics to classify mild dehydration (Sadler, 1997). Therefore, it is possible that there wasn’t a significant water loss for the instrument to detect.

Figure 4.11: Results from Outdoor Dehydration Experiment Conducted on Runners

To promote appreciable fluid imbalance, the study is repeated on the men’s indoor track team. The (five) volunteers ran indoors and were not permitted to drink any fluids two hours prior to practice. Data collection followed the same protocol as before. The results of the study are plotted in Figure 4.12. Although more significant changes occur in the peak to peak voltage potentials, the athletes did not experience enough weight loss to be considered dehydrated by clinical standards. Thus, a definitive statement cannot be made as to the effectiveness of the device.
Testing with athletes concluded because it was difficult to conduct a controlled experiment. Sweat rate and composition is highly variable both between individuals and within an individual (Sadler, 1997). Different rates and compositions imply very different BIA results (high water loss causes a decrease in voltage potential while a loss of both salt and water imply an increase in voltage potential). Additionally, the thermo-response of this elite group of runners may be so well attuned that they efficiently compensate for any water loss, making them less vulnerable to dehydration.

### 4.6.4 Sauna

A sauna is the next method used to try to induce water imbalance. The subject sat in a sauna, operating at 0% humidity and at an approximate temperature of 95°F. BIA
readings, along with the volunteer's weight are noted every fifteen minutes. Weight measurements are taken using a highly precise (+0.1 lb) Detecto scale. To promote water loss, the volunteer refrained from food and drink two hours before testing. To obtain additional information, BIA readings are taken from arm to arm and from arm to foot. Although the results (see Figure 4.13) show similar trends in voltage potential for the two methods of measurement (which is what one would predict), overall, a clear, explainable trend does not emerge.

![Figure 4.13: Sauna Dehydration Test](image)

However, the 3% weight loss (required to be clinically considered dehydrated) is not achieved. Without an independent hydration reference it is uncertain if the person was even mildly dehydrated during the testing. So like before, a conclusive statement cannot be made about the functionality of the device.
The experiment is repeated using the same volunteer, only this time to assure dehydration, testing is to continue until a 3% weight loss is observed. However, water loss via the use of a sauna was found to be an extremely slow process. After five hours of testing only 1.4% drop in body weight is achieved. Without any further appreciable changes in water loss, testing is concluded. The results, (plotted in Figure 4.14) show a decrease in voltage potential as body weight is lost. This trend is the opposite of what is obtained in the first sauna study and contradicts published studies that predict an increase in voltage potential due to water loss. Although, since significant water loss isn’t achieved, the basic validity of the results is in question. Clearly, the hydration instrumentation cannot properly be assessed until water imbalance is certain.

Figure 4.14: Repeated Sauna Dehydration Test
4.6.5 Diuretics

As a final measure, a hydration study is conducted using diuretics. Typically prescribed as medication for edema or high blood pressure, diuretics promote fluid imbalance by directly acting on the kidneys to increase the rate of water and sodium excretion. The resulting depletion in blood volume is compensated for by the movement of extracellular water into the bloodstream (Dulbecco, 1997).

The volunteer is given 40mg of Lasix, a prescribed diuretic. Upon intake, weight and BIA measurements are periodically taken. Weight loss was found to occur linearly with time, at a rate of one pound per hour. The results, shown in Figure 4.15, make intuitive sense. As fluid volume is lost, the body's internal resistance increases, causing a corresponding increase in voltage potential. The shape of the curve is thought to be a function of the body's regulatory response. The body's internal feedback mechanism eventually acts to curb the imbalance so fluid loss does not proceed unbounded.

Although the diuretic study does provide interesting results, the prescribed 3%-5% body weight loss is not achieved. Without an independent hydration reference, a definitive assessment cannot be made about the functionality of the device.
4.7 Summary & Future Work

In this chapter fundamental research was conducted to explore a new BIA methodology. An experimental set-up was built to validate the proposed technique and a variety of mechanisms were employed to disturb the subject’s water balance in a controlled and measurable manner. However, without an independent hydration measurement, one cannot be assured that water imbalance occurred. Without this certainty, the results obtained from the validation tests are inconclusive.

Clinical studies, conducted on individuals diagnosed with fluid disorders, are needed so a definitive assessment can be made about the instrumentation’s capability to detect changes in hydration.
Chapter 5
Discussion & Conclusions

5.1 Summary of Work & Future Considerations

To accommodate the needs of a growing aging population an intelligent robotic aid is being developed by MIT. Intended to assist the elderly living in Assisted Living Facilities, the system facilitates mobility by providing physical support and guidance to the user. Demonstrating the overall concept and enabling technologies, the success of the first test-bed, the SmartCane, prompted the development of the SmartWalker. While differing in structure, the second experimental device is to have health monitoring capabilities.

To adequately determine which health parameters to incorporate into the walker, the user’s capabilities and medical needs needed to be understood. This was accomplished by consulting various geriatric medical professionals. Their input was used to generate a list of indicators of health. Due to time, financial, and system constraints it was ultimately decided that pulse and hydration sensors would be developed for implementation onto PAMM.

With the widespread commercial availability of affordably priced pulse monitors, recording the user’s heart rate was initially viewed as a straightforward task. However, after evaluation they were found to be incompatible with PAMM. Thus, a completely
noninvasive, robust, pulse sensor was developed specifically for the project. Because of
its immunity to mobility disturbances, an ECG based monitor was designed. Detected at
PAMM’s handlebars, the user’s voltage potential is processed by a circuit that filters and
amplifies the waveform. The output is then converted from the analog signal into an
actual heart rate via a peak detection algorithm. Tested rigorously, the system was found
to perform well in conditions similar to those in an assisted living facility.

As a final measure, the integrated pulse monitor should be tested in an actual
Assisted Living Facility with the proposed users. Once it is validated with field trials, the
data obtained from the sensor could provide some interesting research, investigating how
an individual’s heart rate responds to their level of activity. Presumably, this information
could be used as a feedback signal to evaluate user acceptance and the overall
effectiveness of the parameters of the control system.

As one ages, the body’s regulatory mechanisms becomes impaired causing seniors
to be particularly susceptible to fluid imbalance. Because of its profound effect on one’s
health, a hydration monitor was selected for incorporation into PAMM.

The only noninvasive, quantitative tool available that attempts to measure water
imbalance is Bioelectrical Impedance Analysis. The fundamental theory behind BIA is
that the electrical impedance of the human body is a strict function of how hydrated the
person is. Therefore, by introducing a small, high frequency current into the body and
then measuring the resulting voltage potential, an individual’s TBW can be computed.
Due to inherently erroneous assumptions, the current BIA methodology has been shown
to yield inaccurate body water estimates. Therefore, a new, qualitative technique was
proposed. Let a (significant) change in voltage potential from a previous recording suggest a water imbalance.

An experimental device, including the necessary circuitry, was designed and built to properly evaluate the proposed methodology. The underlying intent of the validation experiments was to disturb the subject’s water balance in a controlled and measurable manner to see if the hydration readings would reflect a change. For this to be shown, it first had to be proven that the unit was sensitive enough to even detect fluid imbalance. Therefore, initial testing was conducted on a simple, passive system, beef. Eventually (after many refinements to the experimental procedure), the tests demonstrated the instrument’s sensitivity to changes in water balance. With the assurance of this vital capability, validation testing began with human volunteers. A variety of mechanisms were employed to induce water loss in a safe and observable manner. However, none of the means seemed to accomplish this requirement. Without the financial or medical resources to conduct independent hydration measurements (i.e. isotope dilution or urine analysis), the degree of water lost during testing was unknown. Without the assurance of a fluid imbalance, a conclusive statement cannot be made about the functionality of the device.

Clinical studies, conducted on individuals diagnosed with fluid disorders, are needed so a definitive assessment can be made about the instrumentation’s capability to detect changes in hydration.
References


Godding, S., Field Tests on a Personal Mobility Aid for the Elderly, B.S. Thesis, Department of Mechanical Engineering, Massachusetts Institute of Technology, Cambridge, USA 1999.


Appendix A

Evaluation of Existing Pulse Monitors

Pulse monitors determine when the heart beats by detecting variations at the skin level in one of three parameters: heart sounds, pulsatile blood volume, and surface potentials. Each approach was evaluated for compatibility with the PAMM system.

Heart sounds are vibrations resulting from the closing of the heart valves and the reverberation of the surrounding walls (Guyton, 1976). These vibrations are transmitted along the arteries at the velocity of the pulse wave. Although most of the cardiac vibratory spectrum is below the audible threshold, a microphone designed to detect low-frequency sound waves can be used to sense and amplify these sounds (Norton, 1982). A microphone based pulse monitor was evaluated for implementation. The heart rate is calculated when the tip of his index finger is placed over a sensor pad. Although the monitor is compact and noninvasive, it was found to be incompatible with PAMM because of its vulnerability to disturbances. Reliable readings are only obtained when the user sits perfectly still with his finger fixed on the sensor pad. Even walking while taking a reading leads to large errors. Therefore the monitor is eliminated from future consideration.

As the heart beats, blood flows in and out of compartments in the body. This causes tissue blood volume to change in a pulsatile fashion. Photoelectric sensors and piezoelectric transducers are typically used to detect these changes in volume. Because
of their sensitivity, low-cost, and simplicity, both types of monitors were investigated for possible use with PAMM.

A commercial infrared photoelectric sensor is evaluated. It requires a user to place his finger in a spring clip. An infrared light source and a photo-detector are embedded on opposite sides of the clip. Light is absorbed by the blood, and so as blood volume changes, the amount of light absorbed also changes. Since the electrical signal from the detector is a function of the amount of light falling on it, signal variations are indicative of a heart beat. Although the technique is straightforward and the hardware is compact, the device is not compatible with PAMM. Photoelectric sensors are susceptible to disturbances, even the slightest movement results in false readings. Since there are innate disturbances to this system (a motorized walker moving on uneven surfaces, the unsteady nature of elderly person’s hands) the sensor is not considered further.

Changes in blood volume can also be detected using piezoelectric sensors. Piezoelectric materials develop an electrical current when mechanically stressed (Norton, 1982). When placed under a person’s finger, the volume change caused in the skin due to the flow of blood results in the displacement of the piezoelectric element. Therefore, an electrical current, characteristic of a pulse wave is generated. Piezoelectric film (see Figure A.1) is used to demonstrate this concept.
To obtain a clear pulse signal, the material is placed on a person’s wrist. As shown in Figure A.2, the preliminary results look promising. Although noisy, the pulse wave is recognizable. Since the sensor needs to detect the pulse in an area that comes in contact with the walker, readings are taken at the user’s palm and fingertips. Even though various mechanisms are designed to increase the sensitivity of the film, the material wasn’t sensitive enough to sense the heart beat in these areas.

A commercial piezoelectric disk, fabricated specifically for recording pulse waves at the fingertip, is tried. As shown in Figure A.3, it yields a clean pulse signal. Unfortunately, like the microphone and photo-electric based sensors, it too required the user to remain still while readings were taken. Any sort of movement corrupts the signal. Although various damping techniques, digital filtering and statistical methods are employed to reduce the sensitivity of the piezoelectric disk, the efforts are without success.
Since the commercially available pulse sensors were found to be incompatible with the walker, a robust ECG-based sensor (described in Chapter Three) was created specifically for PAMM.
Appendix B

Heart Rate Monitor Circuitry

Figure B.1: Schematic of Pulse Filter Board
Figure B.2: Hand Detector Circuitry
Appendix C

Hydration Instrumentation Circuitry

Figure C.1: Signal Generator Circuit
Appendix D

Preliminary Meat Experiments

The purpose of these tests is to see if the instrumentation is sensitive enough to detect a water imbalance. A simple, passive system, beef is used. If a known change in water content cannot be sensed in this simple system then it surly cannot be detected in the complex, human body.

A significant water imbalance is imposed on the system by hydrating and then dehydrating the meat sample (see Figure D.1). Changes in water volume are quantified by fluctuations in the piece’s weight. The meat is hydrated by submerging it in a water marinate. The sample is removed from the bath every fifteen minutes so hydration and weight measurements could be taken. Once a 5% increase in body weight is observed, the meat is dehydrated in a self-constructed dryer. The drying mechanism is made similar in function to those used to preserve foods, the critical intention being to desiccate the food, not cook it. The sample is removed from the dryer every fifteen minutes so hydration and weight readings can be recorded. The experiment concluded when a 10% decrease in body weight is noted. For consistency in composition, top round cuts of meat are used.
Conducted three times, the results of the tests are plotted in Figure D.2 and Figure D.3. Although slight trends begin to form, further, more controlled experiments are needed before conclusions could be made.
The testing methodology is examined to reduce errors introduced by the experimental technique and by the instrumentation. Upon consideration, the hydration of the meat was done in a haphazard way, readings were taken without knowing when or if the entire sample was fully saturated with water. A saturation time study is conducted to properly account for this. Three (approximately) equally sized pieces of meat are submerged in water; their weights are recorded every fifteen minutes. As shown in Figure D.4, the saturation time constant is found to be about fifteen minutes with a maximum percent change in weight (for this sample) of 2-3%. Dissimilar degrees of saturation (in any direction) can also lead to erroneous readings. To reduce the gradient in the vertical direction, future experiments are to only include very thin (3/4") sections of beef.

Figure D.3: Results from Dehydration of Meat Sample
The home built dryer was also found to contribute to the unexpected results. To accelerate the evaporative process, in theory the dryer was to dehydrate the meat by elevating the surrounding temperature. However (even at the modest temperature of 80°F) the dryer was found to be (unevenly) cooking the meat. This is undesirable because it alters the chemical and mechanical structure of the sample (which invalidates hydration readings). Establishing an effective drying technique that would not damage the meat's cellular structure was challenging. Various methods were considered: a commercial vacuum cleaner, a vacuum chamber (with reduced atmospheric pressure), a heated vacuum chamber (with reduced atmospheric pressure) and natural environmental conditions (control). Because of spoilage concerns, the effectiveness of each of the methods was quantified by their drying rates. As noted from Figure D.5, although the
heated vacuum chamber is the numerical winner, it is only marginally better than the others.

Figure D.5: Rates of Various Drying Mechanisms

The instrumentation is also examined for possible data contaminates. The electrodes used for data collection were fairly crude. To obtain readings, wires with exposed metal tips were vertically inserted into the meat’s surface; an action which caused a pool of water to surround the tip (See Figure D.6). This arrangement is found to generate a charge distribution at the electrode-electrolyte interface, creating an unwanted capacitive effect (Geddes, 1972). This effect is eliminated with the use of surface electrodes.
Additionally, since the wires were not shielded, the signals interfered with one another. Thus, hardware is built to electronically isolate and mechanically constrain the electrodes (see Figure D.7).

Using the new methodologies and instrumentation, the initial experiment is repeated on a consistently thin, lean piece of top round meat. Like before, weight and BIA
measurements are periodically taken while the sample is hydrated and then dehydrated. However, this time the piece is desiccated by placing it in a heated vacuum chamber. To promote complete equilateral saturation, the specimen is wrapped in saran-wrap before readings are recorded. The modifications to the protocol proved to be effective. As shown in Figure D.8 and in Figure D.9, the results drastically improved. However, the observable trends contradict those predicted by published findings. An excess of fluid should lead to a decrease in resistance (and in voltage potential) while dehydration should cause an increase in resistance (and in voltage potential). This experiment shows the opposite trend. Clearly, other factors are at play.

![Graph showing results from Hydration of Meat Sample]

Figure D.8: Results from Hydration of Meat Sample
Two simple tests are performed to better understand the physics of the system. Voltage potentials are recorded after each sample is cut in the horizontal direction and in the vertical direction, respectively. The results, plotted in Figure D.10 and in Figure D.11 show an increase in potential as the width of the sample decreases, and a decrease in potential when the length of the sample is reduced. This linear relationship is characteristic of a resistive element (inversely proportional to cross sectional area and directly proportional to length). An additional test is conducted to confirm this result. Once again, voltage potentials are recorded for various sizes of meat samples, only this time the frequency of the electrical current is varied from 5-60kHz. Shown in Figure D.12, the response is completely flat, indicative of a resistive element.
Appendix D: Preliminary Meat Experiments
Figure D.12: Frequency Response of Various Sizes of Meat Samples

Once again, to eliminate possible data contaminants, experimental methods are further refined. A scale with improved resolution (to ±0.01g) is used, and since the heated vacuum chamber is only marginally superior to the control, a new drying mechanism is created. As shown in Figure D.13, the set-up relies heavily upon convection effects. To provide heat, a lamp is inserted in the bottom of an aluminum foil lined crate. To circulate the air, a fan is placed over holes punched in the top of the crate. As the meat heats up, moisture filled air rises and is expelled into the environment by the fan. The constant movement of air maintains a low temperature and facilitates the evaporative process, thus providing a highly effective means to dehydrate the sample (without cooking it).
One inherent procedural error was also realized. During the hydration process the meat sample is submerged in a water marinate. This action was done to ensure adequate water balance throughout the entire piece. However, proper saturation can only be achieved if the sample is soaked in a solution of similar chemical composition to that of the body. By placing it in a pure water solution, a fluid imbalance (and erroneous BIA readings) is almost assured. Since extracellular and intracellular fluids are largely (82%) composed of NaCl, a saline solution of appropriate osmolarity is to be used as a marinate in future testing (Guyton, 1976).

As described in Chapter 4, these procedural changes proved to be key. When the experiment is repeated, the expected results are obtained.