STOCHASTIC ANALYSIS OF MOTOR-CONTROL STABILITY, POLYMER BASED FORCE SENSING, AND OPTICAL STIMULATION AS A PREVENTIVE MEASURE FOR FALLS.

by

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ABSTRACT

Falls are the leading cause of all external injuries. Outcomes of falls include the leading cause of traumatic brain injury and bone fractures, and high direct medical costs in the billions of dollars. This work focused on developing three areas of enabling component technology to be used in postural control monitoring tools targeting the mitigation of falls. The first was an analysis tool based on stochastic fractal analysis to reliably measure levels of motor control. The second focus was on thin film wearable pressure sensors capable of relaying data for the first tool. The third was new thin film advanced optics for improving phototherapy devices targeting postural control disorders. Two populations, athletes and elderly, were studied against control groups. The results of these studies clearly show that monitoring postural stability in at-risk groups can be achieved reliably, and an integrated wearable system can be envisioned for both monitoring and treatment purposes.

Keywords: electro-active polymer, ionic polymer-metal composite, postural control, motor control, fall prevention, sports medicine, fractal analysis, physiological signals, wearable sensors, phototherapy, photobiomodulation, nano-optics.
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# TABLE OF CONTENTS

Approval .............................................................................................................. ii  
Abstract .............................................................................................................. iii  
Acknowledgements ........................................................................................... iv  
Table of Contents ............................................................................................... v  
List of Figures ................................................................................................... vii  
List of Tables ..................................................................................................... ix  

## 1: Introduction .................................................................................................... 1  
1.1 Thesis Objectives and Research Contribution ................................... 2  
1.1.1 Summary of Research Objectives................................................... 4  
1.2 Organization of Thesis ................................................................. 4  

## 2: Background and Methodology ...................................................................... 6  
2.1 Physiology of Postural-Control: sensory-motor control systems  
and fall susceptibility .......................................................................... 6  
2.2 Stochastic Analysis of Fractal Systems .............................................. 9  
2.3 Photobiomodulation and Phototherapy Devices ............................... 15  
2.4 Research Methodology .................................................................... 17  

## 3: Clinical testing .............................................................................................. 20  
3.1 Study 1: Hockey Players .................................................................. 20  
3.1.1 Study Design ................................................................................ 20  
3.1.2 Protocol ......................................................................................... 21  
3.1.3 Data Acquisition and Analysis ....................................................... 24  
3.2 Study 1 Results: Hockey Players ..................................................... 25  
3.3 Study 2: Young and Elderly .............................................................. 27  
3.3.1 Study Design ................................................................................ 27  
3.3.2 Subjects ........................................................................................ 27  
3.3.3 Protocol ......................................................................................... 27  
3.3.4 Data Acquisition and Analysis ....................................................... 28  
3.4 Study 2 Results: Young and Elderly ................................................. 29  
3.4.1 Stabilogram Diffusion Analysis...................................................... 29  
3.4.2 Detrended Fluctuation Analysis .................................................... 31  

## 4: Medical device system development .......................................................... 34  
4.1 Force Sensor Development ................................................................... 34  
4.1.1 Electroactive polymer selection and modelling ......................... 34  
4.1.2 IPMC modelling................................................................. 40
LIST OF FIGURES

FIGURE 1: ILLUSTRATES THE SIMILARITIES IN FRACTAL GEOMETRY OF BROWNIAN MOTION (25) AND THAT OF THE CENTRE-OF-PRESSURE MOVEMENT (POSTURAL SWAY) DURING SIMPLE QUIET STANCE OF A HUMAN. .............................................................................................................10

FIGURE 2: LOG-LOG PLOT OF A STABILOGRAM-DIFFUSION ANALYSIS SHOWING MEDIOLATERAL SWAY (X), ANTERIOR-POSTERIOR SWAY (Y) AND THE RESULTANT OF THESE TWO (R). ......................................................12

FIGURE 3: SIMPLE OPEN LOOP VS. CLOSED LOOP CONTROL SYSTEMS DIAGRAM SHOWN WITH RESPECT TO HIGH LEVEL PHYSIOLOGICAL SYSTEMS INVOLVED .................................................................................................................................14

FIGURE 4: SINGLE LEG STANCE TEST .................................................................................................................................21

FIGURE 5: FORCE PLATFORM .......................................................................................................................................................22

FIGURE 6: TEST SUBJECT STANDING ON FORCE PLATE ........................................................................................................23

FIGURE 7: SDA LOG-LOG PLOT SHOWS CLEAR SIGNIFICANCE IN HURST COEFFICIENT. THE HIGHER SLOPE VALUE OF THE PB SUBJECTS CORRESPONDS TO INCREASED LEVELS OF OPEN-LOOP CONTROL (I.E. INCREASED INSTABILITY) ..........................................................................................................................26

FIGURE 8: ANOVA RESULTS SHOWING STATISTICALLY SIGNIFICANTLY DIFFERENT SHORT-TERM HURST COEFFICIENTS MEASURED FROM THE TWO STUDY GROUPS ..............................................................................................................26

FIGURE 9: SDA PLOT OF REPRESENTATIVE ELDERLY SUBJECT’S MEDIOLATERAL, ANTERIOR-POSTERIOR AND RESULTANT SWAY. ........................................................................................................................................29

FIGURE 10: SDA PLOTS OF A REPRESENTATIVE YOUNG SUBJECT’S MEDIOLATERAL, ANTERIOR-POSTERIOR AND RESULTANT SWAY. ........................................................................................................................30

FIGURE 11: DETRENDED FLUCTUATION PLOT: REPRESENTATIVE ELDERLY SUBJECT ................................................................................33

FIGURE 12: DETRENDED FLUCTUATION PLOT: REPRESENTATIVE YOUNG SUBJECT. .................................................................................................................................33

FIGURE 13: TYPICAL CHEMICAL STRUCTURES OF PERFLUORINATED ION EXCHANGE MATERIAL (NA+ CAN BE REPLACED BY OTHER CATIONS SUCH AS Li+ OR H+) ........................................................................................................................................35

FIGURE 14: A DIAGRAM SHOWS THE STRUCTURE OF AN IPMC STRIP AND DEMONSTRATES THAT ANIONS ARE BONDED TO THE BACKBONE (54). .........................................................................................................................36


FIGURE 16: INVERTED IPMC FILM SENSOR RESPONSE FOR POSITIVE DISPLACEMENT INPUT (55) .................................................................38

FIGURE 17: REPRESENTATIVE STRESS-STRAIN CURVE FOR IPMC. MODULUS OF ELASTICITY CALCULATED TO BE 290 MPa. INSET SHOWS A TEST SAMPLE. ......................................................................................................................45

FIGURE 18: A) THE CHARGE INDUCED BY MECHANICAL STRESS ALONG WITH THE TIP DEFLECTION B) MODEL PREDICTION PLOTTED OF CURRENT PRODUCTION ........................................................................................................................................47

FIGURE 19: A) THE CURRENT INDUCED BY VARIOUS TIP DEFLECtIONS UP TO 10 CM, B) CHARGE INDUCED BY VARIOUS DISPLACEMENTS UP TO 10 CM. ....................................................................................................................47

FIGURE 20: 80 MV SIGNAL GENERATED FROM A SINGLE SMALL 2CM TIP DISPLACEMENT IN A CANTILEVER CONFIGURATION. ......................................................................................................................................................49

FIGURE 21: SENSING OF REPEATED DISPLACEMENTS AT RANDOM INTERVALS (~100 Hz). ..............................................................................50

FIGURE 22: LEFT MOST FIGURE SHOWS IONIC EAP WITH GOLD-CHROMIUM ELECTRODE LAYER. RIGHT FIGURE SHOWS MINOR MICRO-CRACKS ON THE SURFACE OF ELECTRODE, LIKELY FORMED DURING THE INITIAL FABRICATION PHASE. .......................................................................................................................50

FIGURE 23: 650 NM SPACED HOLES DESIGNED TO TRANSMIT RED LIGHT AND 420 NM SPACED HOLES DESIGNED TO TRANSMIT BLUE LIGHT. ARRAYS AS SEEN THROUGH AN OPTICAL MICROSCOPE IN DARK FIELD TRANSMISSION ........................................................................................................................................52

FIGURE 24: EXPERIMENTAL SET-UP FOR SEM/FIB DUAL BEAM SYSTEMS. ..................................................................................................53

FIGURE 25: NANOHOLE ARRAYS USED TO ENHANCE PHOTON SIGNAL TRANSMISSION, FABRICATED USING FOCUSED ION-BEAM EMISSION AT SFU. TOP ARRAYS SHOW THE RESULTS OF SLIGHT ASTIGMATISM.
IN THE BEAM CAUSING DEFECTS IN THE HOLES. BOTTOM ARRAY ILLUSTRATES A GOOD EXAMPLE OF A FINELY TUNED BEAM FOR MILLING. ................................................................. 56

FIGURE 26: EXPERIMENTAL SET-UP FOR TRANSMISSION CHARACTERIZATION, OPTICAL MICROSCOPE ON THE LEFT OF THE PHOTO AND SPECTROSCOPE WITH CCD CAMERA ON THE RIGHT ...................... 57

FIGURE 27: NANOHOLE ARRAY TRANSMISSION RESULTS FOR Au-PDMS ARRAYS. DOTTED LINE REPRESENTS THE TRANSMISSION RESULT FOR AN ARRAY DESIGNED TO HAVE A 450NM PEAK. SOLID LINE REPRESENTS THE TRANSMISSION RESULT FOR AN ARRAY DESIGNED TO HAVE A 550NM PEAK. DASHED LINE REPRESENTS THE TRANSMISSION RESULT FOR AN ARRAY DESIGNED TO HAVE A 650NM PEAK. THE TWIN PEAK SEEN THE DASHED LINE IS TWO PEAKS SUPER-IMPOSED AS A RESULT OF TRUNCATING DURING THE SPECTROSCOPE DATA CAPTURE ......................................................... 59
LIST OF TABLES

TABLE 1: COMPARISON AMONG IPMC, SMA AND EAC (53).................................................................37
TABLE 2: OUTLINE ADVANTAGES AND DISADVANTAGES OF BOTH ELECTRONIC AND IONIC EAP
  CATEGORIES ........................................................................................................................................39
TABLE 3: CALCULATED WAVELENGTHS AND WOOD’S ANOMALIES FOR VARIOUS PERIODICITIES AND
  MATERIALS FOR A THIN GOLD FILM (100NM THICK)........................................................................58
1: INTRODUCTION

Postural control, even during quiet stance, requires both feedback and feedforward motor controls to maintain balance. From a neurophysiological motor control point of view, these two strategies are very complex. This is due to the number of systems involved, namely the visual, vestibular and somatosensory, each consisting of a large number of interconnected sub-systems. Measuring specific information from these neural muscular mechanisms is not straightforward, and interpreting those measurements is even more complicated.

One of the most popular traditional methods of measuring postural control is to collect force signals of centre of pressure (COP) data as a time-series. This offers an easy and convenient way with which to examine postural control. However, difficulty lies in interpreting the information contained within these pressure-force traces as they appear to be random and fractal by nature (20). Stochastic systems methods have become a recent area of focus to analyze this fractal nature of complex physiological time series signals, and have been applied to heart rate and blood pressure signals with some success (21) (22). This powerful approach has also been recently applied to centre of pressure time-series in a similar manner (4). The overriding hypothesis of the current work is that this methodology can apply as a reliable predictor of motor control loss. The objective of this work is shown this can be achieved even with low resolution
COP data, making it a suitable candidate for incorporation into wearable devices using flexible force sensors for monitoring and photo-based therapy devices.

1.1 Thesis Objectives and Research Contribution

The initial motivation for this thesis research was two-fold. The first area of focus was to investigate a potential method for reliably predicting postural control loss. Such a method would have to be suitable for integration into an existing wearable photobiomodulation based treatment device. The second area of focus was to research and develop an alternative to current optics for the wearable treatment device.

Typical analysis of complex physiological time-series requires large amounts of data averaged over relatively large time scales making real-time analysis impossible. Stochastic analysis uses non-deterministic statistical techniques to measure the level of randomness occurring in temporal space without the need for such large amounts of data. Algorithms based on stabilogram diffusion analysis and detrended fluctuation analysis were developed using Mathworks Matlab software to determine the stochastic nature of postural motor control using COP data sets. A number of human trials were conducted in order to collect postural control data from a wide variety of subjects including young healthy and elderly subjects, and athletes with problematic balance, in order to test the algorithms specificity. In parallel to the clinical data collection materials were investigate to be used as force sensors to relay COP data in a wearable device. A number of electroactive polymers were researched and a
kind of ionic polymer metal composite (IPMC) was chosen for further investigation. Several sets of IPMC force sensors were designed, fabricated and tested for this purpose.

Phototherapy or photobiomodulation devices and research are considered to be in the early development stage, although it is widely believed that this area will become more prominent in the future for healthcare. One particular device, which has been developed at the Vancouver based company NeuroKinetics Health Services (BC), Inc., has shown promise in a wide array of clinical patients suffering balance dysfunction (2) (38). The device is applied as a shoe insole and stimulates the soles of the feet via a proprietary optical system. This system includes a composite of photo-optical filters to supply specific wavelengths of light target at nerve endings in the foot. A significant limitation of this device is that it requires a number of optical filters stacked in parallel which reduces the transmission power of some wavelengths resulting in uneven treatment dosages for the desired set or range of wavelengths. Another limitation of this device is control over the size of the desired treatment area. This particular device aims to supply treatment to specific low-threshold mechanoreceptors in the glabrous skin of the foot ideally requiring the device to target small groups of afferent neurons. Neurons range in size from 4\(\mu\)m to 100\(\mu\)m in diameter, while the current device exposes an area of 2cm\(^2\). A replacement to the optical filter component allowing better control over photo transmission and offering scalability down to the micron level is desired.
1.1.1 Summary of Research Objectives

1. Develop a methodology and algorithm for fast, more accurate, and reliable monitoring and analysis of postural-control during normal standing conditions, such as quiet stance.

2. Design a robust force-sensor that could be used in conjunction with the above as an integrated centre-of-pressure postural-control monitoring system.

3. Design a controlled nano-optical filtering device to produce and expand upon the optical parameters set out by the partner company.

1.2 Organization of Thesis

In chapter 2 the motivation of this research is explained through a brief exploration of the physiological systems being measured and what role the methodology and technology developed in this work may play in future biomedical research. With this background knowledge, the reader is also given an overview of the research methodology that was used to direct and ultimately complete this thesis.

In chapter 3 the design of the clinical trials are explained. The subject protocols and how they evolved from the first to the second study are described in respect to the lessons learned from the first study. How the data from each study was collected and analyzed is described in detail.

In chapter 4 the components for a medical device to measure and treat postural control deficiencies are described. Selection of an appropriate material for force sensing was necessary and a number of potential materials are listed and reviewed. Modelling, fabrication and testing of the selected ionic-metal
composite material is discussed. The design, testing and fabrication of the proposed nano-optical filtering system is described in detail.

In chapter 5 the results of the two clinical studies and the performance results relative to the two component technologies are detailed.

In chapter 6 the results in chapter 5 are discussed in detail and analyzed with respect to a working medical device for postural control monitoring and fall prevention.

In chapter 7 the conclusions of this thesis and possible avenues for future work are explored.

Appendices A, B, and C outline the Matlab scripts used to complete this research. Appendix D illustrates several centre-of-pressure force platform plots (stabilograms).
2: BACKGROUND AND METHODOLOGY

2.1 Physiology of Postural-Control: sensory-motor control systems and fall susceptibility

Humans are able to maintain a posture that is perfectly adapted to the surroundings thanks to a central neural multimodal organization making up the principle components of the sensory-motor control system (1). The sensory-motor controls are the core neural system that provides balance control to the human body during static and dynamic performance. Key components of the sensory-motor controls include the visual-ocular system, vestibular system, the oculomotor and paravertebral muscles, foot mechanoreceptors, and limb proprioception. All which play a part in maintaining and adapting the posture to the surroundings as well as monitoring a person’s movement and motion in space. These sub-systems coordinate postural neuromuscular tonus to handle and naturally optimize body positioning and balance control during static and dynamic activities (2).

The components giving rise to functional joint stability must be flexible and adaptable to meet the requirements of varied tasks. The process of maintaining functional joint stability is accomplished through a complementary relationship between static and dynamic components, and is of particular concern for the elderly. Ligaments, joint capsule, cartilage, friction, and the bony geometry within the articulation, comprise these static components (3) (4). Dynamic contributions arise from feedforward and feedback neuromotor control over the skeletal
muscles crossing the joint. Also contributing to the effectiveness of the dynamic restraints are the biomechanical and physical characteristics of the joint, such as range of motion and muscle strength and endurance.

For the elderly, postural disturbances can be introduced externally or internally in daily living and frequently impose the threat of a fall. To maintain standing balance, the ability to detect postural disturbances and generate proper postural responses is a major requirement. This ability has long been noted to deteriorate with age leading to decreased balance and heightened risk of falling (5) (6), although an agreement on the cause for this have remained elusive. The most promising evidence is observed in the deterioration in the function of the somatosensory and motor systems which appears to occur with age, and has been found to be related to poorer static standing balance (7) (8). In addition to maintaining balance while performing daily activities, the ability to counteract unexpected externally induced forces (reactive postural control) is also essential for independent living. However, in addition to the elderly, altered reactive postural responses are also been observed in healthy young adults who are experimentally deprived of normal somatosensory function, and in patients with vestibular dysfunction or peripheral neuropathy (9) (10). Limited contraction capacity in the gastrocnemius, and smaller muscle response amplitudes to support surface perturbations, have also been observed in functionally unstable older adults compared with stable older adults (11). Therefore, the control of reactive balance is also under the influence of the function of the sensory-motor systems.
Recent studies investigating postural stability with the elderly have revealed several key parameters in predicting fall susceptibility, including time-series analysis of mediolateral and anterior-posterior sway, area analysis of total centre of pressure plot (stabilogram), and velocity of sway (12) (13) (14). Of these parameters, the time series created from mediolateral direction of sway is perhaps the most important in identifying elderly fallers (15) (16) (17). This time series is easily captured using standard force plate tools, and therefore holds the potential for easy non-invasive methods of fall prediction. Measuring these time-series while suppressing visual inputs (e.g. having the subject close their eyes) has shown to produce the most consistent results differentiating balance conditions and predicting fallers. In one very comprehensive study, subjects with suffering multiple falls within a 16 month period, had significantly more mediolateral sway measured during balance control tests when their visual inputs were removed (17). The reasoning for this is clear, by removing the visual sensory-motor control input the subject must rely solely on the reaming sensory-motor inputs. In the case of quite standing, the measurements taken are directly related to the vestibular system and foot mechanoreceptors, and to a lesser extent limb proprioception.
2.2 Stochastic Analysis of Fractal Systems

A stochastic system or process is one whose behaviour is non-deterministic. This means that the system’s subsequent states will be determined by both the processes’ predictable actions and by a random element(s). The word stochastic is derived from the Greek term for ‘to aim’ or ‘guess’. A fractal refers to an object or quantity that displays self-similarity that can be scaled infinitely up or down. One of the most common and easily imagined examples of a fractal is the geometry of a snowflake.

A unique component of this research has been the use of stochastic (non-deterministic) and fractal (non-linear) analyses to increase the understanding of postural control with clinical applications in mind. Clinical diagnosis and basic investigations are critically dependent on the ability to record and analyze physiological signals, however the analysis of such signals yield little in reliable information. Often, this is due to an absence of characteristic temporal and/or spatial scales that may confer important biological advantages (18) (19). This challenge can be overcome with more sophisticated stochastic and fractal analyses which do not require such temporal or spatial scales. Fractals are the results from stochastic processes. Another fractal example is Brownian motion, the seemingly random movement of particles suspended in fluid. It is this particular example that can be used analogously to the motion produced by the postural sway of a human, as illustrated in the figure below.
One of the most developed techniques in stochastic analysis of COP data was first developed in the 1990’s and is known as stabilogram diffusion analysis (SDA) (23) (12) (24) (25). A stabilogram refers to the plot of COP created from the force plate data. This method models COP data as a correlated random-walk, and the short-term and long-term processes are calculated via a set of six parameters. These parameters are expressed as the critical point coordinates (separating the dynamics into short- and long-term regions), and for each region a diffusion coefficient. These parameters are realized from Einstein’s classical law for Brownian motion (26) and were defined first for COP by Collins and De Luca (23).
Equations (1.1) and (1.2) represent the mean square planar displacement between all pairs of the entire centre of pressure time series, separated by the timer interval $\Delta t$.

\[
\langle \Delta x^2 \rangle \approx 2D\Delta t \quad (1.1)
\]
\[
\langle \Delta x^2 \rangle = \Delta t^{2H} \quad (1.2)
\]
\[
\langle \Delta x^2 \rangle_{\Delta t} = \frac{\sum_{i=1}^{N-m} (\Delta x_i)^2}{N - m} \quad (1.3)
\]
\[
C = 2(2^{2H-1} - 1) \quad (1.4)
\]

Plotting $\Delta x$ as a function of $\Delta t$ (equation (1.3)) yields the stabilogram diffusion plot where the diffusion coefficient is the calculated slope(s) of the two regions. The slope of the log-log plot is the Hurst coefficient ($H$), and is shown in the below figure, where $H$ can be any real number in the range of $\{0<H<1\}$. The correlation function for fractional Brownian motion, $C$, is time –dependent and is given by equation 1.4 (27).
As can be seen from equations (1.2) and (1.3) one feature of fractional Brownian motion is that the past increments in a particle’s displacement are correlated with the future increments; the only exception being when $H = 0.5$ which corresponds to a classical random walk (i.e. classical Brownian motion). It is necessary to note that when $H>0.5$ the stochastic process is positively correlated (i.e. $C>0$). In this case the Brownian movement in a particular direction for some time, $t_0$ will tend to continue in the same direction for $t > t_0$. A general statement can now be made: increasing trends in the past indicate on the average increasing trends in the future, and decreasing trends in the past indicate on average decreasing trend in the future. In this particular case when
H>0.5, a higher value of the Hurst exponent implies an increase in the level of randomness of the system, where H=1 is a purely random system.

The opposite occurs for H<0.5, where past and future increments are negatively correlate (C<0). Here increasing trends in the past imply decreasing trends in the future and decreasing trends in the past imply increasing trends in the future. In this case, a lower value of the Hurst exponent implies an increase in the level of randomness.

Analysis of a COP time series in this manner defines higher and lower levels of stochastic activity that can be inferred as strategies of either open loop and closed loop control respectively (23) (24). In the study of motor control, a generalization can be made: open loop control corresponds to a less ordered sensory-motor control strategy and vice versa for closed loop control (28). Figure 3 illustrates the differences between open and closed loop control strategies, effectively closed loop control allows for optimization by using feedback from the environment and open loop does not. The great advantage of SDA is it provides a direct analogue of the mathematical result to the motor control status of an individual relative to a reference measurement. For example, if H>0.5 for two statistically different subject groups, the group exhibiting the greater value of H could be said to have a higher level of stochastic activity in their postural control and thus be exhibiting characteristics of an open loop control strategy. The significance of these components and their analogous value to motor control been verified and are well detailed in various texts (23) (29) (30).
A complimentary technique to stabilogram diffusion analysis is detrended fluctuation analysis (DFA). DFA is another stochastic systems technique that obtains an exponent similar to the Hurst exponent, except that DFA may also be applied to signals whose underlying statistics (such as mean and variance) or dynamics are non-stationary (31). DFA provides a more reliable fractal analysis method in long-term temporal intervals, where SDA tends to be less reliable (12). In DFA an alpha coefficient (the slope of the line produced from the calculated regression line) is measured, however, from analysis of the alpha coefficients the Hurst coefficient may be calculated and is thus useful for comparison with SDA.
If alpha values are greater than 0.5 then $H=1-\alpha$, if they are less than 0.5 then $H=\alpha$. In this project, DFA was used for calculation and analysis of long-range Hurst coefficients, that is for longer-term intervals. Analysis of SDA has suggested that over the short-term intervals open-loop control schemes are more identifiable while over long-term intervals closed-loop control mechanisms become more prevalent. Since the focus of this research is to identify when individuals postural control is waning information on the prevalence of open-loop control is of more interest and therefore there is a greater emphasis in this work on the short-term interval SDA results. Initial research coupling the two methods has received an exciting level of success in the analysis of postural control (32) (12).

2.3 Photobiomodulation and Phototherapy Devices

The use of low levels of visible or near infrared light for reducing pain, inflammation and edema; promoting healing of wounds, deeper tissues and nerves; and preventing tissue damage by reducing cellular apoptosis has been known for almost forty years since the first development of lasers. Originally thought to be a peculiar property of laser light (soft or cold lasers), the subject has now broadened to include photobiomodulation and photobiostimulation using non-coherent forms of light. Despite many reports of positive findings from experiments conducted in vitro, in animal models and in randomized controlled clinical trials, photobiomodulation remains somewhat controversial. This likely is due to the complexity of choosing amongst a large number of illumination
parameters such as wavelength, fluence, power density, pulse structure and treatment timing that has led to the publication of a number of both positive studies (mostly human clinical trials) and negative ones (33).

Several studies have reported at least temporary symptomatic reversal of diabetic peripheral neuropathy during treatment with monochromatic near-infrared photoenergy (MIRE) delivered non-invasively by the Anodyne® Therapy System (ATS; Anodyne Therapy LLC, Tampa, FL 33626, USA) (34) (35) (10). The studies on MIRE have shown promise, reporting both negative and positive results, but have been limited only to controlled human clinical trials.

Several very recent studies have demonstrated cellular chromophores responsible for the effect of visible light on mammalian cells, including cytochrome c oxidase (36) (37), with absorption peaks in the near infrared, and photoactive porphyrins. Mitochondria are thought to be a likely site for the initial effects of light, leading to increased ATP (adenosine triphosphate) production, modulation of reactive oxygen species and induction of transcription factors. These signal transduction pathways in turn lead to increased cell proliferation and migration (particularly by fibroblasts), modulation in levels of cytokines, growth factors and inflammatory mediators, and increases in anti-apoptotic proteins. The results of these biochemical and cellular changes in animals and patients include such benefits as increased healing in chronic wounds, improvements in sports injuries, pain reduction in arthritis and neuropathies, and amelioration of damage after heart attacks, nerve injury and retinal toxicity.
2.4 Research Methodology

Research into the area of human balance has shown that reliable predictors of falls may be obtained by monitoring changes in mediolateral centre of pressure (COP) data from force-plates (14). In the study of fall susceptibility and prevention, measures of postural control are the primary method towards a quantification of sensory-motor function. The force platform is the most widely used tool to measure postural control parameters. However, there exists some controversy as to which parameters are most effective in the prediction of fall susceptibility. A recent comprehensive literature review of fall related studies conducted in elderly subjects from 1950 to 2005 identified several specific parameters most likely to be associated with fall susceptibility (14). The best indicators were mean speed and amplitude of mediolateral centre of pressure movement in eyes open and eyes closed conditions, as well as the root mean square value of mediolateral displacement of the centre of pressure. Measures of dynamic posturography (moving platforms) were not indicative of predicting falls (15) (14). After careful review of these previous studies, an AMTI AccuSWAY force plate to measure and collect postural movement data was chosen for this project.

The issue of monitoring these measurements with a wearable device (such as a specialized shoe or shoe insert) was addressed in the first phase of this research. The main challenge was to create and test an algorithm based on the SDA/DFA mathematics that could reliably and rapidly measure and analyze the motor control using relatively low-resolution data (~100 Hz) collected from a force sensor small and light enough to be comfortable for the wearer that does
not impede the individual’s balance. The resolution of data was chosen in order to reduce computation time necessary for the analysis while maintaining a high level of reliability. For this purpose a kind of electro-active polymer (EAP) was chosen to be developed as a potential sensor. If it is possible to develop a reliable model of “prior-to-fall” perturbations typically seen, an EAP-based system could be used as a force sensor to sense postural abnormalities and actuate imbedded therapeutic devices or warning systems. EAP properties make them particularly useful in biomedical applications (39).

The second phase of the proposed research is to combine the preventative ability to predict potential loss of balance with a mitigating technology to stimulate and correct the reduced postural control. The objective here was to validate the transmission spectrum of the NeuroKinetics device and look at other means to capture the same set of photonic parameters while address the two limitations discussed earlier in section 1.3. Addressing these issues will allow controlled stimulation to only the primary mechanoreceptors that relay proprioceptive and tactile information necessary for postural control from the lower limbs in the foot (40) and thereby eliminating overstimulation over surrounding nerve tissue (33).

From a number of technologies the concept of sub-wavelength nano-holes seemed to meet all the criteria: controlled stimulation of areas down to 4 µm in diameter, scalable in size, high reliability in the peak wavelengths being transmitted, as well as very low losses in photon power (41). Sub-wavelength nanohole arrays enhance the transmission of light when illuminated through the coupling of surface plasmon polaritons (SPP). SPPs are essentially
electromagnetic waves trapped at the surface of a thin metal or conductive film due to an interaction with the metals surface free electrons. The transmitted output of such arrays is dependent on the geometry and periodicity; and therefore high resolution of control on wavelengths transmitted can be achieved. Importantly, this allows for optical very good tunability without suffering losses seen in standard optical devices.

The nano-optics required were developed at SFU’s 4D laboratory Nanoimaging facility. Nanohole arrays were fabricated using a FEI 235 dual-beam focused ion beam (FIB) and imaged with a field emission scanning electron microscope to produce 100nm to 200nm diameter sub-wavelength holes in 100nm thick gold films, ranging in size from 1µm² to 10 µm².
3: CLINICAL TESTING

3.1 Study 1: Hockey Players

3.1.1 Study Design

The first study’s goal was to collect data in order to analyze motor control strategies of individuals identified with problematic balance control and to test the feasibility that even in small populations SDA can differentiate between these two groups where standard summary statistics fail to do so. The results of this study were presented at the 2008 Western Canadian Conference on Environmental Ergonomics and Physiology in Burnaby, British Columbia (42). Eleven male and one female hockey players were assigned into either a poor balance group (PB, n=6) or control (CON, n=6) group based on their performance on a timed single-leg stand (43) (44). This test was then compared to the results from a simple force-plate test. Force-plate measurements of centre of pressure sway were recorded at 100Hz for sixty seconds during quiet standing with eyes closed. Summary statistics were computed to assess overall balance performance between the PB and CON groups, while Stabilogram Diffusion Analysis (SDA) was used to assess differences in motor control strategies.
3.1.2 Protocol

3.1.2.1 Pre-screening Tests

Balance performance was assessed using two tests: right and left single-leg balance, and an extreme balance board. These two tests are regularly used by trainers and team physicians to assess balance performance in hockey players (43) (44). For the first test the subject stood stationary on the right leg similar to that shown in Figure 4. The subject’s eyes were closed and they were asked to hold their knee in their hands for support. This position was held for a count of ten breathes (the player inhaled and exhaled ten times). The task was repeated three times for each leg and an average score determined from number of breathes taken until loss of balance occurred was recorded.

Figure 4: Single leg stance test
3.1.2.2 Force Plate Tests

Postural sway data was collected and processed using an AMTI AccuSwayPLUS Force Plate (Figure 5) provided by NeuroKinetics. Postural sway data was recorded at a rate of 100 Hz via eight analog channels through a fourth order Butterworth lowpass filter. The voltage signals from the analog channels were calculated to represent mediolateral (“side to side”), anterior-posterior (“back and forth”) and downward directional forces of the subjects. Examples of the calculations are shown below, where \( F_x \) represents ML sway; \( F_y \) represents AP sway; \( F_z \) represents downward force; A, B, C, XDC etc. represent the analog channels; S11, S21 etc. are matrices provided by the manufacturer AMTI to convert the voltage signals into units of force:

\[
F_x = (\Delta C_z * S_{11}) + (\Delta D_z * S_{12}) + (\Delta A_z * S_{13}) + (\Delta B_z * S_{14}) + (\Delta YAC * S_{15}) + (\Delta XDC * S_{16}) + (\Delta XAB * S_{17}) + (\Delta YBD * S_{18}) \\
(2.1)
\]

\[
F_y = (\Delta C_z * S_{21}) + (\Delta D_z * S_{22}) + (\Delta A_z * S_{23}) + (\Delta B_z * S_{24}) + (\Delta YAC * S_{25}) + (\Delta XDC * S_{26}) + (\Delta XAB * S_{27}) + (\Delta YBD * S_{28}) \\
(2.2)
\]

\[
F_z = (\Delta C_z * S_{31}) + (\Delta D_z * S_{32}) + (\Delta A_z * S_{33}) + (\Delta B_z * S_{34}) + (\Delta YAC * S_{35}) + (\Delta XDC * S_{36}) + (\Delta XAB * S_{37}) + (\Delta YBD * S_{38}) \\
(2.3)
\]

---

Figure 5: Force platform
For each balance test, the participants stood quietly from a seated position into a stand with both feet on a force plate for sixty seconds while force plate data (e.g. centre of pressure, forces and moments) were collected. Precautions were taken in case a subject was to fall, safety mats were placed on the floor in front and behind the force plate as well as spotters were positioned to the right and left of the participant during all data collection procedures.

![Test subject standing on force plate](image)

**Figure 6: Test subject standing on force plate**

Participants stood in a closed based foot position on the force platform. Tests were sixty seconds in duration and performed with the participants eyes closed. Participants looked at a simple point target place on the wall in front of them for temporal-spatial reference just prior to initiating each balance test (closing their eyes).
3.1.3 Data Acquisition and Analysis

3.1.3.1 Force Plate Data

From the available data, four variables were selected for basic statistical analysis (15):

- $X_{Davg}$: Mean displacement of medial lateral sway (cm)
- $Y_{Davg}$: Mean displacement of anterior-posterior sway (cm)
- $A_c$: Circular area of centre of pressure plot (cm$^2$)
- $V_{avg}$: Average velocity of sway in mediolateral and anterior posterior directions (cm/sec)

Examples of COP plots of force plate data are shown in Appendix D.

It has been shown in previous works that mean displacement and velocity in the mediolateral direction are the most reliable indicators of difficulties with balance and susceptibility to fall (14). The reason for this is likely due to the geometry of the lower limbs. The ankle (tibiotarsal) joint is, for example, mainly a simple hinge joint, allowing rotation (plantar-flexion/dorsi-flexion) in the sagittal plane. Hence, from a passive mechanical standpoint, upright bipedal stance is considerably more stable in the frontal plane than in the sagittal plane (30).

The summary statistics were compared with stabilogram diffusion and detrended fluctuation analyses.
3.2 Study 1 Results: Hockey Players

Analysis of the force plate summary statistics showed no significant difference between the control (CON) and problem balance (PB) groups of subjects in mediolateral ($p = 0.49$) and anterior-posterior directions ($p=0.64$).

The SDA analysis showed PB to have a significantly higher ($p=0.0029$) average Hurst coefficient ($H= 0.867\pm0.022$) compared with CON ($H=0.815\pm0.024$) in the mediolateral direction, see Figures 18 and 19. No significant difference in the anterior-posterior direction. Compared to the single leg balance test, SDA analysis of force plate data differentiated between CON and PB subjects very well considering the small size of the subject groups. The SDA results imply that the subjects in the poor balance group use a predominant open-loop strategy for postural control while standing. The motor control strategy identified in the PB group may be interpreted as a greater tendency to move away from a relative equilibrium point over the short term, thus causing problematic balance control. The use of SDA provided an important analysis tool in the assessment of the neural mechanisms associated with loss of balance control. The next step was to validate the generalizability (25) of this approach in using populations of healthy young and elderly subjects.
Representative SDA Plot:

Figure 7: SDA log-log plot shows clear significance in Hurst coefficient. The higher slope value of the PB subjects corresponds to increased levels of open-loop control (i.e. increased instability).

ANOVA for Groups:

Figure 8: ANOVA results showing statistically significantly different short-term Hurst coefficients measured from the two study groups.
3.3 Study 2: Young and Elderly

3.3.1 Study Design

The second study followed directly from the first, young subjects (n=8) and elderly subjects (n=4) were tested using a protocol built on from the first study using the same AMTI AccuSWAY force plate used in the previous study. Two methods were done to analyze the force plate data to test for the most reliable fractal analysis for fall analysis: Stabilogram Diffusion Analysis (SDA) and Detrended Fluctuation Analysis (DFA) were done.

3.3.2 Subjects

Seven young (24.3 ± 3.1 years) and five elderly (69.2 ± 4.3) healthy subjects participated in the study. Informed written consent was obtained from the subjects prior to each experiment. The study conformed to the standard set by the Declaration of Helsinki and the experimental protocols were approved by the office of Research and Ethics for Simon Fraser University.

3.3.3 Protocol

All subjects completed one testing session. Within this testing session subjects performed two sit-to-stand tasks (45) randomized with eyes open or closed. The eyes closed protocol met using snugly fitted blindfold to occlude vision. Upon arrival to the laboratory, each subject was asked to remove his or her shoes and sit in front of the force plate (46). The subjects were given instruction on the protocol prior to experimentation. They were told to stand upright, after an initial 5-minute sitting period, to a pre-determined foot position. The subject’s foot position was kept constant using markings on the force plate.
that corresponded to the top of the distal phalange of the big toes and the heels of the subject at a shoulder-width stance for each subject. The initiation of the 5-minute standing period was timed from the moment both feet touched the floor when the force plate read the subjects full weight bearing onto the plate. Force data were collected immediately prior to standing and during the full 5 minutes of standing. A 10-minute rest period was given in between trials. In both conditions, subjects were instructed to sit and stand as still as possible with both arms along the body. In the eyes open condition, subjects were instructed to look directly at a target that was fixated at eye-level on the wall. To prevent disturbances of the subject by visual and auditory stimuli, all measurements were taken in a quiet room with no features in the field of view other than the target and the walls enclosing the testing area that were draped with black sheets.

### 3.3.4 Data Acquisition and Analysis

The analog force plate signals were recorded simultaneously at 100Hz using a computerized strip chart recorder. Centre or pressure magnitudes and variation were determined from the force plate from eight analog channels at 100Hz. Force plate data was also sent to specialized acquisition software related to the AccuSwayPLUS (46) with a time synchronization signal from the cardiovascular data.

Summary statistics, SDA and DFA were used to measure statistical differences between the groups.
3.4 Study 2 Results: Young and Elderly

3.4.1 Stabilogram Diffusion Analysis

For SDA the short-range Hurst coefficient were calculated to provide a reliable fractal analysis method in short-term temporal intervals. The young subjects studied exhibited significantly lower Hurst coefficients than the elderly subjects in both the mediolateral and anterior-posterior directions of sway (p = 0.028 and p = 0.007 respectively). The young subjects Hurst coefficients for mediolateral (Hxs: 0.694 ± 0.068) and anterior-posterior (Hys: 0.726 ± 0.09) sway and the elderly subjects (Hxs: 0.812 ± 0.10; Hys: 0.856 ± 0.035).

Figure 9: SDA plot of representative elderly subject’s mediolateral, anterior-posterior and resultant sway.
These results indicate that after the initial period of standing-up right, the young subjects are able to command a closed-loop strategy of motor-control providing a better mode for postural control performance compared to the elderly subjects. The elderly subjects exhibited significantly less statistical chance of closed loop control within the initial period of standing.

An unexpected result was found during the short transition period between sitting and standing. In contrast to the steady state results illustrated in Figures 20 and 21, the elderly subjects exhibited lower Hurst coefficients (Hxs: 0.777 ± 0.11; Hys: 0.845 ± 0.072) than that of the young subjects (Hxs: 0.870 ± 0.02; Hys: 0.890 ± 0.044) in both mediolateral and anterior-posterior sways. The results were right at the cut-off of being statistically different for Hx (p = 0.050) and not significant in for Hy (p = 0.188), however from observation the individual subject results it does appear to be a trend, more subjects and trials would be
required to validate this result. If validated, this result would imply that Younger subjects utilize a greater trend for open-loop motor control during the action of standing. This makes sense as a young healthy individuals would be more likely to commit to a premeditated action with an exact model for execution. The elderly subjects on the other hand, exhibiting the trend of a closed-loop control strategy, are perhaps more likely to proceed in the action with more caution requiring feedback from the environment during the action and never fully committing to the action. This could due to a number of factors including reduced muscle tonus and/or sensory ability, and deserves further investigation with a larger pool of subjects.

3.4.2 Detrended Fluctuation Analysis

For DFA, the long-range Hurst coefficients were measured. DFA provides a reliable fractal analysis method in long-term temporal intervals. In DFA an alpha coefficient (the slop of the line produced from the calculated regression line) is measured, however, from analysis of the alpha coefficients the Hurst coefficient may be calculated from the alpha coefficients (useful for comparison with SDA results). If alpha values are greater than 0.5 then $H = 1 - \alpha$, if they are less than 0.5 then $H = \alpha$.

No significant differences in the long-term temporal range were found. However some comments should be made about these results. While the differences were not significant, the mean values of the Long term Hurst coefficients were found to be small in the elderly than the young subjects in both the AP and ML directions. This finding is similar to that of Collins and De Luca’s
using different stochastic techniques (30). It suggests that the steady-state behaviour of the closed-loop postural control mechanisms in the elderly is more negatively correlated and therefore perhaps more stable over the longer term (of steady state). There is an increased probability that the movements away from a relative equilibrium point will be offset by corrective adjustments back towards the equilibrium point. This change in control dynamics from the short-term could be due to an increase in the gain of the feedback mechanisms that are involved in the regulation upright posture. A postural control strategy of this sort may allow elderly individuals to compensate for alterations in steady-state behaviour of the open-loop control mechanisms, offsetting the effects of an increased tendency for short-term drift (as seen in the SDA results). These changes in the long-term dynamics of the postural control system could account for the finding that the long-term effective Hurst coefficients for the healthy elderly were smaller, though not significantly so, than those for the young subjects. This could be interpreted as an increase in gain decreasing the probability of the COP moving a larger distance away from the equilibrium point or shifting between different points, and thus leading to lower long-term effective Hurst coefficients. The differences between the young and elderly long-term effective Hurst coefficients may not have been significant because the proposed gain changes would have had to offset the indirect effects of the elderly subjects' increased tendency for short-term drift. Therefore, it is expected that an increase in short-term drift or fluctuations across the multiple joints of the body would lead to an increase in the long-term fluctuations of the overall system.
Figure 11: Detrended Fluctuation Plot: Representative Elderly Subject

Figure 12: Detrended Fluctuation Plot: Representative Young Subject
4: MEDICAL DEVICE SYSTEM DEVELOPMENT

4.1 Force Sensor Development

4.1.1 Electroactive polymer selection and modelling

Electroactive polymers (EAP) are those polymers which either respond to an external electrical input by displaying a property change, such as size or shape displacement, or inversely respond to a property change by outputting an electrical signal. EAPs are lighter and more flexible than other sensing or actuating materials such as shape memory alloys (SMA) or electroactive ceramics (EAC); the striction capacity of EAPs can be several orders of magnitude higher than EACs. EAPs are also generally low cost and have fewer fabrication steps allowing for easy creation of application specific sizes and shapes. Furthermore many kinds of EAPs are biocompatible and the operate at low electric-potentials making them particularly interesting for biomedical applications (47) (48) (49) (50) (39).

EAPs may be classified into two categories: Electronic EAP and Ionic EAP. The former group behaves actively to the external electric field due to the Coulomb forces generated by the built-in charges or dipoles, while the dominating driven mechanism of the later group is the diffusion of ions (47).

Several ionic EAP have been reported in the literature, for example, Polypyrrole (PPy) [2]. An ionic polymer-metal composite (IPMC) is chosen due to its commercial availability and maturity of the relevant technologies. The
following is a brief review of materials and technology investigated for this projects application in force sensing.

4.1.1.1 Ionic Polymer-Metal Composites

IPMC is a biocompatible material (51) and is based on ion exchange polymers (IEP). IEP refers to the polymers which selectively exchange ions of a single charge (either cations or anions) with their own incipient ions. Standard IEP consists of fixed covalent ionic groups, for example, Perfluorinated alkenes with short side-chains terminated by ionic groups (SO$_3^-$ or COO$^-$) for cation exchange or ammonium cations for anion exchange Figure 7, and Styrene/divinylbenzene-based polymers where the phenyl rings have been substituted by ionic groups as shown in Figure 7. The former group has short side-chains, providing ionic groups to interact with solvent molecules and control the passage of appropriate ions, while the later group is highly cross linked and rigid (52). Therefore, a perfluorinated alkene ion exchange material, Nafion™ (e.g. Nafion 115 or 117) of DuPont was picked in this project for its commercial availability and relatively well defined material properties. Moreover, as a commercially available product, Nafion offers better uniformity than other lab-made IPMC (51).

![Figure 13](image_url): Typical chemical structures of perfluorinated ion exchange material (Na$^+$ can be replaced by other cations such as Li$^+$ or H$^+$) (53).
Cations inside the hydrated Nafion membrane can move freely within the polymer matrix while the anion maintains a bond to the fluorocarbon backbone (Figures 7 and 8) (54). The type of the mobile cations inside IPMC affects the sensing/actuating performance. Extensive evaluation work proves that small cations, like Li+, can offer a better sensing/actuating performance (52).

Figure 14: A diagram shows the structure of an IPMC strip and demonstrates that anions are bonded to the backbone (54).
Table 1 compares the IPMC with SMA and EAC in terms of their sensing properties. These results indicate that IPMC offers excellent reaction speeds with a robust and less dense material. It is more flexible and the reaction speed is acceptable in most biomedical applications.

<table>
<thead>
<tr>
<th>Property</th>
<th>IPMC</th>
<th>SMA</th>
<th>EAC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Force (MPa)</td>
<td>10-300</td>
<td>~700</td>
<td>30-40</td>
</tr>
<tr>
<td>Reaction speed</td>
<td>μs to s</td>
<td>s to min</td>
<td>μs to s</td>
</tr>
<tr>
<td>Density</td>
<td>1-2.5 g-cm⁻³</td>
<td>5-6 g-cm⁻³</td>
<td>6-8 g-cm⁻³</td>
</tr>
<tr>
<td>Fracture toughness</td>
<td>Resilient, elastic</td>
<td>Elastic</td>
<td>fragile</td>
</tr>
</tbody>
</table>

Table 1: Comparison among IPMC, SMA and EAC (53).

If an IPMC membrane is bent mechanically, the applied stress will contract one side while spread the other one. A gradient of charge concentration is consequently set up and the mobile cations will move toward the region with lower charge density (Figure 9). Since the anions are fixed, this process will result in the excess of positive charges in the expanded side and the deficit in the compressed side, generating an electrical potential. If a short circuit is configured, a current will be produced (52) (55). The sensing result can be transformed from the short circuit current into a voltage by using a current-to-voltage converter.
Figure 15: Scheme of IPMC working as a motion sensor: the arrow shows the cation moving with the direction of the mechanical bending force (52).

Most reported sensing voltages are less than 100 mV and can be easily sensed by a low-power amplifier (52). As a sensor, IPMCs do not require working within a hydrated environment. For an IPMC strip with cantilever configuration, a linear relationship between the sensing voltage and the tip displacement has been reported (52) (53) (56). IPMC sensors are able to respond to a dynamic load up to hundreds of Hz as well, which expands its application even further (51).

Figure 16: Inverted IPMC film Sensor response for positive displacement input (55).
Dielectric EAP, Ferroelectric Polymers, Electrostrictive Graft Elastomers, Electrophoretic Paper, Electro-Viscoelastic Elastomers and Liquid Crystal Elastomer (LCE) Materials are many examples of electronic EAPs. Ionic EAPs include Ionic Polymer Gel (IPG), Ionomeric Polymer-Metal Composites (IPMC), Conductive Polymers (CP) and Carbon Nanotubes (CNT). The advantages and disadvantages of Electronic EAP and Ionic EAP are summarized in Table 2 (47) (39).

<table>
<thead>
<tr>
<th>EAP Type</th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
</thead>
</table>
| Electronic | • Can operate in the air  
• Can hold strains under DC voltage  
• Induce relatively large actuation forces  
• Respond faster | • Require a higher voltage  
• Require compromise between strain and stress  
• Require a higher temperature  
• Glass transition temperature is high |
| Ionic | • Require low voltage  
• Provide predominately bending actuation  
• Exhibit large displacement  
• Quick Response  
• Work well in low frequencies | • Most of them cannot hold strain under DC voltage  
• Bending EAP has a relatively low actuation force.  
• Can be complex fabrication process |

Table 2: Outline advantages and disadvantages of both electronic and ionic EAP categories

Electret polymers are polymers that have a quasi-permanent electric charge or dipole polarization. The electret polymer layer can store charge in a similar way to dielectric layers inside a capacitor. The difference is that the polarization in the capacitor dielectric is only transient, while the electret polymer exhibits a quasi-permanent charge storage or dipole polarization. Some of
electret polymers also display ferro-electricity. These materials are in the thermodynamic equilibrium and can hold the polarization permanently (57).

PVDF is one of the most common types of polar polymers, based on the dipole polarization. EMFi, is a cellular polymer that has built-in permanent charges and is also very common. Generally, both of them have ferroelectric-like properties and they are biocompatible, flexible and easy to use. EMFi is less dense and has lower Young’s modulus than the polar polymers. Due to its cellular structure, it also has a larger piezoelectric tensor element that is due to the large deformation of the maroscopic dipoles under mechanical load. On the other side, PVDF has higher Pyroelectrocity and stronger longitudinal piezoelectric response. The major difference in property between these two is the polarity of piezoelectricity. The cellular polymers have positive intrinsic piezoelectricity while the piezoelectric coefficients of the polar polymers are opposite (dipole-density piezoelectricity) (58). The biomedical sensors based on either PVDF or EMF are commercially available in the current market. However, due to the requirement of high driving voltages in comparison with ionic EAP along with greater weight and cost of manufacture make them unsuitable for this application.

4.1.2 IPMC modelling

The models discussed in the current work were produced in MATLAB to validate their results. These previous models were built-on in order to improve IPMC modelling and appropriate selection and design for this particular application.
There have been several good attempts by different research groups to model IPMCs for force sensing. While these groups have taken different approaches to modelling IPMCs, there are several shortfalls that were identified to be addressed and built on:

1) All models have used Euler-Bernoulli beam theory to model mechanical stiffness and deformation. This model assumes small deformations (and strains) and tightly constrains the geometry of the beams.

2) All models are currently based around applications requiring high frequency sensing (~1000 Hz). There has been some recent evidence that IPMCs may respond best at lower frequencies (56).

3) Most recent models assume dry (i.e. non-hydrated) IPMC specimens although dry IPMC specimens are not often adequately characterized for the mechanical properties. All models use different values of Young’s modulus that they have measure themselves using variable (and sometimes questionable) techniques.

4) None of these models account for large deformations (strains), or for oddly shaped geometry, and no empirical measurements have been recorded in the literature for such variables.

5) The methods in which all models are built assume that voltage and current are directly related to the mechanical and geometrical properties of the IPMC, however no recorded work has been reported which extends these models to device designs that optimize these features in order to
produce optimal voltage and current outputs. This is likely because these models make too many assumptions in their gray or black box approaches and are unable to handle and significant modifications.

6) All models consider a simple free-fixed-end cantilever beam. No models have attempted to model other configurations such as 3-point bends, torsion etc.

Previous efforts by other groups’ have concentrated on actuation and sensing properties of IPMCs (51). While previous models have shown a direct relationship between electric charge production and mechanical/geometrical properties, numerous assumptions and a black or grey box approach has limited them in usefulness for investigation of specific features such as power storage. After reproducing several previous models several reoccurring assumptions were identified that did not lend well to modelling storage of charge (59) (55) (56) (60). It was these assumptions that were targeted for improvement on previous models. The first assumption previous models made were using Euler-Bernoulli beam theory to model mechanical stiffness and deformation in a cantilevered configuration. The drawbacks to this theory are in the assumption of only small strains and tight constraints on geometry. The second assumption of previous models was operations only in the high frequency ranges (100-1000Hz) although it has been very recently shown in 2007 that IPMCs respond best at lower frequencies for sensing and power generation (59). The third assumption for these models are dry ionic polymer membranes, however dry IPMC specimens
have not yet been adequately characterized for their electromechanically properties in previous works. Most groups have used mechanical properties (Young’s modulus etc.) of hydrated IPMCs for their model inputs, while others characterizing their own samples with no standardization of the mechanical testing techniques (ASTM etc.) used between groups.

The model presented here utilizes and builds on previous efforts by Farinholt et al (61) and is used to estimate the maximum theoretical charge produced due to mechanical deformation in IPMCs of varying geometry and mechanical properties. Here we show how this is used as the basis for estimating the energy density storage potential of a given IPMC in order to model the sensing characteristics of a given device. First, the mechanical stress, $\sigma$, measured in pascals (Pa), and charge density, $\rho$ are related through the generic term, $\psi$, as shown:

$$\sigma = \psi \rho \quad \text{(1)}$$

Then standard beam theory of moments is applied to give the stress at a given point as:

$$\sigma = -\frac{3Yh(L-z)w_L}{L^3} \quad \text{(2)}$$

Where $Y$ is the equivalent Young’s modulus of the composite, $w_L$ is the tip deflection, $h$ is the thickness of the beam, $L$ is the unsupported length of the beam, and $z$ is the distance along the length from the tip where the force is applied. The equivalent Young’s modulus is calculated using the rule of mixture (62) from the following micromechanics formula:
\[ Y = Y_{EL} V_{EL} + Y_{NF} V_{NF} \]  \hspace{1cm} (3)

Where \( Y_{EL} \) and \( V_{EL} \) are the Young’s Modulus and percentage volume of the electrode, \( Y_{NF} \) and \( V_{NF} \) are the Young’s Modulus and percentage volume of Nafion respectively. This is an important step as the value of the Young’s Modulus is significantly reduced if values for Nafion alone are used alone. As can be seen clearly from the following equations, the value of Young’s Modulus is directly related to charge density and current. These values for equivalent Young’s modulus were confirmed through mechanical characterization tests (using ASTM D 882) of our fabricated IPMCs (see Figure 11), along with stable ion exchange ability, it has been concluded that Nafion has excellent mechanical properties.
Figure 17: Representative stress-strain curve for IPMC. Modulus of Elasticity calculated to be 290 MPa. Inset shows a test sample.

Solving for $\rho$ by substituting in equation 1 into 2 and using the Young’s Modulus from equation 3, we have the initial charge density at any point along the length of the polymer:

$$\rho = \frac{3Y}{\psi} \frac{h(L-z)}{L^3} w_L$$  \hspace{1cm} (4)

Solving for the charge and current using equation 4 we have:

$$q(t) = w_L \frac{3Ybh}{2\beta \psi L} e^{-2\nu}$$  \hspace{1cm} (5)
\[ i(t) = -\lambda wL \frac{3Ybh}{2\beta \psi L} e^{-\lambda t} \]  \tag{6}

Where \( \lambda \) is defined as a time constant and \( \beta \) is a function of \( \lambda \) (59), the width of the beam is denoted by \( b \), and \( t \) is the time in seconds.

Using these equations as a basis for charge density and current which may be contained in a given IPMC membrane, the charge and discharge characteristics may be estimated by assuming either an initial voltage or current input using the following:

\[ i(t) = \left( \frac{V_\infty - V_0}{R} \right) e^{-t/\tau} \]  \tag{7}

Where the \( V_\infty \) is the input voltage and \( \tau \) is the time constant, equal to the capacitance, \( C \), multiplied by the resistance, \( R \). The time constant can be solved for from equation 7 and the electromechanical characteristics of an IPMC modelled. We found this model had excellent agreement with experimental tests, see Figure 11.

The plots below were produced using experimental data, for a beam 2.5 cm x 0.85 cm and 180 \( \mu \)m thickness, with a Young’s modulus of 290 MPa, with a maximum tip displacement of 0.2 cm.
The following plots show the results of this model for a beam 15 cm x 6 cm and 200 µm thickness, and a Young’s modulus of 600 MPa (63) subjected to a 10 cm tip displacement using the same model. The model predicts a linear relationship and that larger and stiffer (higher Young's modulus) will yield a much larger charge and current production, particularly under larger deflections.

Figure 18: a) the charge induced by mechanical stress along with the tip deflection b) model prediction plotted of current production.

Figure 19: a) the current induced by various tip deflections up to 10 cm, b) charge induced by various displacements up to 10 cm.
4.1.3 IPMC fabrication and testing

Electrodes were fabricated on a 1 cm x 5 cm sample of Nafion 115 (DuPont) with thickness of 127µm, consisting of 100nm of Chromium (Cr) and 300nm of Gold (Au) respectively by RF sputtering in SFU Engineering Science Clean Room to create the ionic polymer metal composite. The chromium is used to ensure good adhesion as gold typically does not adhere well to most materials. It was not known however if adhesion would be maintained. Two wire leads on both sides (“top” and “bottom”) at the same end on the IPMC were applied using conductive silver epoxy. This was done because the sputtered electrodes are too thin to use standard soldering for connections.

The IPMCs were tested using a Tektronix TDS 1001B Digital Storage Oscilloscope for its electromechanical sensing properties. The IPMCs were tested for mechanical and structural properties using an Instron 5848 Microtester using dog bone shapes. The test specimens were prepared as specified by the ASTM standards 638 and 882 (64).

4.2 IPMC Sensing Results

The IPMC was placed in a cantilever configuration, and it was observed that for 2 cm bends, 80µV were produced and reproduced in this sensing configurations. Repeated bends with finger tapping (5 Hz) produced repeated voltage spikes. The figures below show the results of these tests.
Figure 20: 80 mV signal generated from a single small 2cm tip displacement in a cantilever configuration.
One of the concerns with such thin electrodes are micro-crack propagation occurring over time reducing the conductivity and therefore sensing ability of the IPMC force sensor. Another concern is the bonding of the electrode to the surface of the polymer. Gold does not adhere well to tightly woven molecular surfaces, such as that found with polymers, which is why a thin layer of chromium is first sputtered on to the polymer to provide an adhesive layer to bond the two materials together. Scanning electron micrograph was performed to investigate these two concerns. It was found that even after nine months the electrodes continued to exhibit excellent bond strength to the polymer surface. Some micro-cracks were observed using SEM analysis although did not appear to affect the electrical conductivity of the electrode.

Figure 22: Left most figure shows ionic EAP with gold-chromium electrode layer. Right figure shows minor micro-cracks on the surface of electrode, likely formed during the initial fabrication phase.
4.3 Photobiomodulation development

4.3.1 Nano-array modelling and design

Demonstration of enhanced transmission through periodic arrays of subwavelength holes in optically thick metallic films has been of recent interest in the development of smaller more sensitive sensors (65) (66). Not only is the transmission much higher than expected from classic diffraction theory, it can actually be greater than the percentage area occupied by the holes, implying that the light impinging on the conductive material between the holes can be transmitted. This means the whole periodic structure acts similar to an antenna in the optical regime, neatly demonstrating the benefits that surface plasmon modes can provide. The transmission spectra of nanohole arrays display peaks that can be tuned by adjusting the period and the symmetry to get results as displayed in Figure 14.
Figure 23: 650 nm spaced holes designed to transmit red light and 420 nm spaced holes designed to transmit blue light. Arrays as seen through an optical microscope in dark field transmission.

The nanohole arrays support enhanced transmission at specific wavelengths that match the surface plasmon resonance conditions. For normal incidence the surface plasmon resonance ($\lambda_{\text{max}}$) can be calculated using the equation:

$$\lambda_{\text{max}}(i,j) = p(i^2 - j^2)^{-1/2} \left( \frac{\varepsilon_d \varepsilon_m}{\varepsilon_d + \varepsilon_m} \right)^{1/2}$$

Where indices $i$ and $j$ are the scattering orders from the array. For a free-standing nanohole array, incident light is diffracted/scattered by the array, producing evanescent waves that tunnel through the holes, resulting in a small but finite amplitude on the far side of the array. Here the evanescent waves are diffracted or scattered, the interference of the resulting waves produces the light that propagates away from the structure. The above equation acts as a starting
point in analysing the transmission spectrum, a spectrum that is more accurately
determined by taking into account these diffraction/interference effects (65).
Surface plasmons act to enhance the fields associated with the evanescent
waves, therefore producing a method of increasing the transmittance. When the
metal film is thin enough, this tunnelling becomes resonant because the SP
modes on the two surfaces can overlap and interact via the holes (65).

Using this equation along with characterization, sets of nanohole arrays
were fabricated to match the transmission spectrum requested by the industrial
partner.

4.3.2 Nano-array fabrication and testing
Nanohole arrays were fabricated using a FEI 235 dual-beam focused ion
beam (FIB) and imaged with the field emission scanning electron microscope
(SEM), at 4D labs located in SFU, shown in the figure below.

Figure 24: Experimental set-up for SEM/FIB Dual Beam systems.
Accurately controlling the etch patterned shape, depth and spacing was achieved with the FIB utilizing various parameters defined within a stream file that the FIB reads and uses for control of the beam and sample stage (67). In the present work, the gallium ion beam was set to 30 keV for milling with a milling rate of 1.6 nmµs\(^{-1}\) for gold and a beam current of 50-200 nA at 5000x magnification. At this magnification the spot size of the FIB is 7.14nm (measured and calibrated using the “spot size” function within the FIB controlling software). Parameters set as such produce 200 nm diameter sub-wavelength holes in 100-nm thick gold films. Periodicities (distances between individual holes within the array) were adjusted as needed within the stream files created for reading by the FIB. The etch depth is primarily dictated by the beam current and dwell times. The beam current controls the intensity of the ion beam and can be defined from within the menu of the FIB interface. The dwell time is simply the amount of time the ion beam scans a particular pixel, the longer the beam scans the more material is etched away. One can control and set these parameters with in the FIB console or they can be predefined in a stream file and read by the FIB once the beams have been properly tuned. A stream file is a series of pixel coordinates compiled into a data file along with the beam dwell time in nano or micro seconds. When read by the FIB console the stream file dictates the x and y coordinates to be scanned by the ion beam. At a magnification of 5000x, the field of view consist of 4096x4096 pixels with each pixel corresponding to 7.14nm. The stream file can be created by a script file to be read as an input file containing the geometry of the structure, including periodicity, as well as dwell
times. Stream files for this work were designed and created using a script written in MATLAB shown in Appendix C.

This technique allows one-step nanofabrication with precise control of hole-size and placement, and the fabrication of highly ordered arrays with well defined geometry. A major advantage of using the dual-beam systems is immediately after the FIB has milled the nano-holes the geometry and quality of the arrays can immediately validated using SE imaging without readjusting the ion beam. Readjusting the ion beam could lead to astigmatism or focusing issues that can drastically influence the quality of the structures being milled, examples shown in the figure below.
Figure 25: Nanohole arrays used to enhance photon signal transmission, fabricated using focused ion-beam emission at SFU. Top arrays show the results of slight astigmatism in the beam causing defects in the holes. Bottom array illustrates a good example of a finely tuned beam for milling.

The photonic properties of the nanohole arrays were characterized using dark field illumination with linearly polarized light on a Zeiss Axio Imager M1m optical microscope. Samples were dispersed on glass microscope slides. Scattered light from the nanostructures was collected with 50× or 100× objectives and analyzed using a PI Acton MicroSpec-2360 spectrometer with a PIXIS.
400BR CCD camera system. The setup is located in 4D Labs (www.4dlabs.ca/) at SFU and is shown in the figure below.

Figure 26: Experimental set-up for transmission characterization, optical microscope on the left of the photo and spectroscope with CCD camera on the right.

4.4 Nano-array transmission results

The wavelengths for the surface plasmon modes and Woods Anomalies were predicted for modes (1, 0) and (0, 1) using equations in section 3.3 and adjusted for the SP resonance shift (41) (65) by modelling and choosing the appropriate periodicity values. Because these equations do not take into account the presence of the holes and their associated scattering losses, they neglect the interference that gives rise to a resonance shift (41). The consequence here is a
prediction of peak positions at wavelengths slightly shorter than those observed experimentally (41). However, this can be accounted for by knowing the resonance shift of an array of given periodicity. A number of arrays with different periodicities ranging from 250nm to 750nm were fabricated using FIB (see figures 15 and 16). The arrays were analyzed individually with a spectroscopy microscope and camera (see Figure 17). Table 3 shows predictions of peak wavelengths and Woods Anomaly calculated using the equations and methods outlined in section 3.3 and taking into account the actual resonance shifts of arrays with the given periodicities measured experimentally. For example, in order to achieve a peak wavelength of ~450nm a thin gold film coated with the transparent flexible organic polymer polydimethylsiloxane (PDMS) coating, holes of the array should be spaced between 250nm and 275nm apart (see Table). These values were used to design the arrays for maximum light transmission for a desired wavelength.

<table>
<thead>
<tr>
<th>Periodicity (nm)</th>
<th>Wavelength (nm)</th>
<th>Woods Anomaly (nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Air-Gold</td>
<td>Glass-Gold</td>
</tr>
<tr>
<td>250</td>
<td>256</td>
<td>383</td>
</tr>
<tr>
<td>275</td>
<td>282</td>
<td>421</td>
</tr>
<tr>
<td>300</td>
<td>308</td>
<td>460</td>
</tr>
<tr>
<td>325</td>
<td>333</td>
<td>498</td>
</tr>
<tr>
<td>350</td>
<td>359</td>
<td>536</td>
</tr>
<tr>
<td>375</td>
<td>385</td>
<td>574</td>
</tr>
<tr>
<td>400</td>
<td>410</td>
<td>613</td>
</tr>
</tbody>
</table>

Table 3: Calculated wavelengths and Wood’s anomalies for various periodicities and materials for a thin gold film (100nm thick).
A typical set of transmission spectrums from arrays with three different periodicities yielding three distinct transmission peaks is shown in Figure 27. Key parameters of the transmission spectrums are noted on the figure.

Figure 27: Nanohole array transmission results for Au-PDMS arrays. Dotted line represents the transmission result for an array designed to have a 450nm peak. Solid line represents the transmission result for an array designed to have a 550nm peak. Dashed line represents the transmission result for an array designed to have a 650nm peak. The twin peak seen the dashed line is two peaks super-imposed as a result of truncating during the spectroscope data capture.
5: ANALYSIS AND DISCUSSION

5.1 Postural Sway Analysis

The results observed in the SDA indicate that after the initial period of standing upright, the young subjects are able to command a closed-loop strategy of motor-control providing a better mode for postural control performance (30) much more so than exhibited by the elderly subjects. While there were no significant differences observed between young and elderly subjects for DFA analysis of postural sway, the mean values of the long-term Hurst coefficients were found to be smaller in the elderly than the young subjects in both the anterior-posterior and medio-lateral directions. This finding is comparable to that of a single previous study done by Collins and De Luca (1995) which used SDA to compare healthy subjects to head trauma patients. Although the results in that study were understandably more exacerbated between the two groups (30) than the current work, the general trend for head trauma patients to exhibit open loop controlling strategies and failing to maintain balanced posture is very similar to that of the elderly subjects studied in the current work. This suggests that steady-state behaviour of the closed-loop postural control mechanisms in elderly are more negatively correlated and therefore more stable in steady state over a longer term. This control strategy could allow the elderly to compensate for alterations in steady-state behaviour of open-loop control mechanisms, offsetting the effects of an increased tendency for short-term drift observed seen in the SDA results.
5.2 IPMC Performance

When a properly prepared IPMC membrane is bent mechanically the applied stress will contract one side and spread to the other side creating a gradient of charge concentration resulting in mobile cations moving toward the region of lower charge density. The sensing voltages measured from the IPMC sensors prepared were on the order of 100mV which could easily be sensed by a low power amplifier. Sensing voltages of this level were consistent a data rates ranging from 20Hz to 100Hz. Unlike previous works these IPMC sensors do not require a hydrated environment. In a cantilever configuration the relationship between the sensing voltage and the tip displacement has been found to linear and scalable in physical size of the sensor.

5.3 Nanohole Technology as Future PBM Devices

Photobiomodulation has been applied to disorders with various forms of light exposure for the treatment of certain disorders or improvement to health and cosmetics. PBM been used for sensory-motor stimulation to improve postural control including in the reduction of tremors in Parkinson’s disease (68) (69) and treatment for diabetic peripheral neuropathy (70) (71). Other disorders which can benefit from phototherapy treatments include acne vulgaris (72), seasonal affective disorder and adjustments in circadian rhythm for delayed sleep phase syndrome (73) (74), and photo-stimulated chemotherapy in cell targeted skin
cancer treatment, and enhanced wound healing (75). Current technology for PBM include the use of optical filters, lasers, light-emitting diodes (LEDs), fluorescent lamps, and other narrow-band, multi-band, or full spectrum lamps. The illumination area varies from exposing small tissue surfaces with focused light to irradiating larger regions from a distance. For example, in treatment of acne vulgaris (commonly known as “severe pimples”), visible violet light in the range of 405nm to 420nm at a total dosage of 320J/cm2 has been demonstrated to effectively destroy the bacteria within the pores and thus effectively treat the condition for 76%-80% of patients (72).

Although optics technology has advanced greatly over the recent years with the increasing popularity and maturity of smaller, more energy efficient, and more reliable silicon LEDs and fiber-optics, many of the equipment used in PBM are still rigid, large, bulky, and at best, table-top portable systems. Due to the rigid nature of the lamp sources, it is very difficult to irradiate small targeted areas of the skin without risking exposure to other sensitive areas for prolonged durations of time. For example, in diabetic neuropathy, patients are required to sit or lie down while infrared light is administered to the soles of the feet and/or legs, and they are thus limited to short durations of time (1-2 hours), there are no wearable devices which allow for longer, “on-the-go” stimulation. A wearable device of this kind would allow physicians and researchers to study physiological performance while under PBM stimulation, target the specific areas needed for stimulation and provide treatment while doing an activity potentially leading to quicker recovery times. These could potentially result in much higher success-
rates due to administration of “better” irradiance-dosages and increased willingness of patients to comply with longer, daily treatment programs.

Nanohole arrays hold the potential to allow the designer to specify the precise wavelengths of interest for a given application, including defining the bandwidths and transmission power. Taking advantage of the surface plasmon resonance of light interacting at the metal surfaces, at high percentage of light can be produced with highly specified parameters. This allows for exact PBM treatments without the losses of standard optical filtering techniques.
6: CONCLUSIONS AND FUTURE WORK

This thesis introduced three necessary proofs of concept towards a wearable device to monitor and mitigate fall occurrence for at risk groups ranging from elderly to high performance athletes. The understanding and prevention of falls is an important area of study that could lead to a dramatic decrease in the leading cause of all external injuries. Athletes, who are often victim to head injuries, have demonstrated acute balance deficits making them more susceptible to falls that may cause further injury. This work was shown that stabilogram diffusion analysis can be used as an effective tool in assessing sensory-motor control recovery in respect to postural control. Future work should concentrate on working with larger populations of subjects and focus on even more specific areas of postural control. One example of this could be focusing on the area of transition between open-loop and closed-loop control strategies and validating the differences between elderly and young subjects as initially identified here.

The feasibility of a wearable device for real time monitoring has been shown through the research, design and development of an ionic polymer-metal composite based force sensor capable of measuring centre of pressure force data. More work on this material is needed to validate it’s life-cycle while in use as a sensor such as high cycle, fatigue and hysteresis testing which could add much to the models presented in the current work.
Treatment or mitigation of postural control loss is a much more difficult area to address. Phototherapy devices such as those being developed at NeuroKinetics hold an enormous amount of potential. The issues of scalability and precise control over photonic parameters such as transmission peaks and bandwidths have been addressed and should lead to further improvement in the design and implantation as a therapeutic device. This device is ready for larger clinical trials to validate its use with specific focus groups.
APPENDICES

Appendix A – Matlab Scripts for Stabilogram Diffusion Analysis.

This script performs a detrended fluctuation analysis on a nonstationary input signal to obtain an estimate for the scaling exponent. The first section is setup to allow input of a standard ASCII text or Matlab file. The script was developed base on original Matlab scripts provided to the author by Andrea Stamp at Boston University. There are three matlab scripts, the first builds a simple GUI interface (Stamp, 2006), the second script is the main function of the SDA and the third script defines a function to traverse the second derivative of the data and finds the times that the linear regression curves should be fitted with.

First SDA Script:

```matlab
h = figure;
set(h,'name','SDA Program');
set(h,'menubar','none');
set(h,'color',[0 0 0]);

analyze = uimenu(h,'Label','File Analyze');
adattocp = uimenu(analyze,'Label','dat -> cp','CallBack','datcp,');
acptodf = uimenu(analyze,'Separator','on','Label','Label','cp ->
df','CallBack','cpdf,');
adattodf = uimenu(analyze,'Separator','on','Label','dat ->
df','CallBack',['datcp',',','cpdf,']);

options = uimenu(h,'Label','Options');
```
oparam = uimenu(options,'Label','Show Parameters','Checked','on','CallBack',
'[if strcmp(get(oparam,"Checked"),"off"),
';set(oparam,"Checked","on"),
'else,
'set(oparam,"Checked","off"),
'end,]);

opoints = uimenu(options,'Separator','on','Label','# of data points');

o3000 = uimenu(opoints,'Label','3000','Checked','On','CallBack',[if
'strcmp(get(o3000,"Checked"),"off"),
'set(o3000,"Checked","on"),
'set(o6000,"Checked","off"),
'end,]);

o6000 = uimenu(opoints,'Label','6000','CallBack',[if
'strcmp(get(o6000,"Checked"),"off"),
'set(o6000,"Checked","on"),
'set(o3000,"Checked","off"),
'end,]);

graphics = uimenu(h,'Label','Display');
greg = uimenu(graphics,'Label','Show Regression Lines','Checked','on','CallBack',[if
'strcmp(get(greg,"Checked"),"off"),
'set(greg,"Checked","on"),
'else,
'set(greg,"Checked","off"),
'end,]);

gx = uimenu(graphics,'Separator','on','Label','X');

gxlin = uimenu(ngx,'Label','Linear','Checked','on','CallBack',[if
'strcmp(get(ngxlin,"Checked"),"off"),
'set(ngxlin,"Checked","on"),
'else,
'set(ngxlin,"Checked","off"),
'end,]);

ngxlog = uimenu(ngx,'Label','Log','Checked','on','CallBack',[if
'strcmp(get(ngxlog,"Checked"),"off"),
'set(ngxlog,"Checked","on"),
'else,
'set(ngxlog,"Checked","off"),
'end,]);

gy = uimenu(graphics,'Label','Y');

gylin = uimenu(gy,'Label','Linear','Checked','on','CallBack',[if
'strcmp(get(gylin,"Checked"),"off"),
'set(gylin,"Checked","on"),
'else,
'set(gylin,"Checked","off"),
'end,]);

gylog = uimenu(gy,'Label','Log','Checked','on','CallBack',[if
'strcmp(get(gylog,"Checked"),"off"),
'set(gylog,"Checked","on"),
'else,
'set(gylog,"Checked","off"),
'end,]);

gr = uimenu(graphics,'Label','R');
grlin = uimenu(gr,'Label','Linear','Checked','on','CallBack',[if
'strcmp(get(grlin,"Checked"),"off"),
'set(grlin,"Checked","on"),
'else,
'set(grlin,"Checked","off"),
'end,]);

grlog = uimenu(gr,'Label','Log','Checked','on','CallBack',[if
'strcmp(get(grlog,"Checked"),"off"),
'set(grlog,"Checked","on"),
'else,
'set(grlog,"Checked","off"),
'end,]);

graw = uimenu(graphics,'Separator','on','Label','Plot Raw CoP data',
'CallBack','plotraw,');
gmult = uimenu(graphics,'Separator','on','Label','Plot Multiple df files',
'CallBack','plotmult,');
fileframe = uicontrol(h,'Style','frame','Position',[0 310 315 110]);
filetext = uicontrol(h,'Style','text','String','Input Options','Position',[35 365 200 50]);
filenum = uicontrol(h,'Style','text','String','# of files:','Position',[5 375 65 20]);
filenum2box = uicontrol(h,'Style','frame','Position',[100 375 50 20]);
filenum2 = uicontrol(h,'Style','edit','String','10','foregroundColor',[0 0 0],'Position',[100 375 50 20]);
filepatho = uicontrol(h,'Style','text','String';'Path to open from:','Position',[5 355 120 20]);
filepatho2 = uicontrol(h,'Style','edit','String','/home/users/','foregroundColor',[0 0 0],'Position',[125 355 185 20]);
filepathc = uicontrol(h,'Style','text','String','Path to save to:','Position',[7 335 100 20]);
filepathc2 = uicontrol(h,'Style','edit','String','/home/users/','foregroundColor',[0 0 0],'Position',[125 335 185 20]);
filedata = uicontrol(h,'Style','text','String','Files:','Position',[7 315 40 20]);
filedata2 = uicontrol(h,'Style','edit','foregroundColor',[0 0 0],'Position',[45 315 265 20]);
**Second SDA Script:**

```matlab
set(h,'Pointer','watch');
datafilesnum = str2num(get(filenum2,'String'));
datapatho = get(filepath2,'String');
datapathc = get(filepathc2,'String');
datafiles = get(filedata2,'String');

if strcmp(get(o3000,'Checked'),'on'),
    NPTS = 3000;
else
    NPTS = 6000;
end

LAGMAX = 1000;
    X = zeros(1,LAGMAX+2);
    Y = zeros(1,LAGMAX+2);
data1 = strtok(datafiles);
for i=1:datafilesnum,
    data = strtok(datafiles);
    load([datapatho,data,'.cp'])
datafiles = strrep(datafiles,strtok(datafiles),' ');
    temp = eval(data);

    [a b] = size(temp);

    tempX = temp(1:NPTS,1);
    tempY = temp(1:NPTS,2);
    tempX = tempX';
    tempY = tempY';
    tempX = tempX - mean(tempX);
    tempY = tempY - mean(tempY);
```
corrX = xcorr(tempX,tempX,'biased');
fix = corrX; %calculate correlations
corrX = corrX(NPTS:NPTS+LAGMAX+1);
corrX = 2*var(tempX) - 2.*corrX;

corrY = xcorr(tempY,tempY,'biased');
corrY = corrY(NPTS:NPTS+LAGMAX+1);
corrY = 2*var(tempY) - 2.*corrY;

X = X + corrX;
Y = Y + corrY;
end

X = X./datafilesnum;
Y = Y./datafilesnum;
R = X + Y;

fprintf('
Writing data to file: %s\n',[data1,'.df']);
 fid = fopen([datapathc,data1,'.df'],'wt');
 Z = [X; Y];
 fprintf(fid,'\n%f %f\n',Z);
 fclose(fid);
 filt = firls(40,[0 1],[1 1],'differentiator');
derdata = filter(filt,1, X);
der2data = filter(filt,1,derdata);
[fitTimeX,critpointX] = findtime(der2data);

derdata = filter(filt,1, Y);
der2data = filter(filt,1,derdata);
[fitTimeY, critpointY] = findtime(der2data);

derdata = filter(filt,1, R);
der2data = filter(filt,1,derdata);
[fitTimeR, critpointR] = findtime(der2data);
\[
\begin{align*}
\tau &= 0.01:.01:(\text{LAGMAX} +2)*.01; \\
\tau_{1X}\text{plot} &= \text{fitTimeX}(1):.01: \text{fitTimeX}(2)+1.2; \\
\tau_{2X}\text{plot} &= \text{fitTimeX}(3)-1.2:.01: \text{fitTimeX}(4); \\
\text{fitcoeffx1} &= \text{polyfit}(\text{fitTimeX}(1):.01: \text{fitTimeX}(2),X(\text{fitTimeX}(1)*100: \text{fitTimeX}(2)*100),1); \\
\text{linex1} &= \text{polyval}(...);
\end{align*}
\]
\[
\begin{align*}
\text{fitcoeffx2} &= \text{polyfit}(\text{fitTimeX}(3):.01: \text{fitTimeX}(4),X(\text{fitTimeX}(3)*100: \text{fitTimeX}(4)*100),1); \\
\text{linex2} &= \text{polyval}(...);
\end{align*}
\]
\[
\begin{align*}
\tau_{1Y}\text{plot} &= \text{fitTimeY}(1):.01: \text{fitTimeY}(2)+1.2; \\
\tau_{2Y}\text{plot} &= \text{fitTimeY}(3)-1.2:.01: \text{fitTimeY}(4); \\
\text{fitcoeffy1} &= \text{polyfit}(\text{fitTimeY}(1):.01: \text{fitTimeY}(2),Y(\text{fitTimeY}(1)*100: \text{fitTimeY}(2)*100),1); \\
\text{liney1} &= \text{polyval}(...);
\end{align*}
\]
\[
\begin{align*}
\text{fitcoeffy2} &= \text{polyfit}(\text{fitTimeY}(3):.01: \text{fitTimeY}(4),Y(\text{fitTimeY}(3)*100: \text{fitTimeY}(4)*100),1); \\
\text{liney2} &= \text{polyval}(...);
\end{align*}
\]
\[
\begin{align*}
\tau_{1R}\text{plot} &= \text{fitTimeR}(1):.01: \text{fitTimeR}(2)+1.2; \\
\tau_{2R}\text{plot} &= \text{fitTimeR}(3)-1.2:.01: \text{fitTimeR}(4); \\
\text{fitcoeffr1} &= \text{polyfit}(\text{fitTimeR}(1):.01: \text{fitTimeR}(2),R(\text{fitTimeR}(1)*100: \text{fitTimeR}(2)*100),1); \\
\text{liner1} &= \text{polyval}(...);
\end{align*}
\]
\[
\begin{align*}
X(1) &= X(1) + .00001; \\
Y(1) &= Y(1) + .00001; \\
R(1) &= R(1) + .00001; \\
\text{logX} &= \log10(X); \\
\text{logY} &= \log10(Y); \\
\text{logR} &= \log10(R);
\end{align*}
\]
fitcoeffx1log = polyfit(log10(fitTimeX(1):.01:fitTimeX(2)),logX(fitTimeX(1)*100:fitTimeX(2)*100),1);
linex1log = (tau1Xplot.^fitcoeffx1log(1))*10^fitcoeffx1log(2);

fitcoeffx2log = polyfit(log10(fitTimeX(3):.01:fitTimeX(4)),logX(fitTimeX(3)*100:fitTimeX(4)*100),1);
linex2log = (tau2Xplot.^fitcoeffx2log(1))*10^fitcoeffx2log(2);

fitcoeffy1log = polyfit(log10(fitTimeY(1):.01:fitTimeY(2)),logY(fitTimeY(1)*100:fitTimeY(2)*100),1);
liney1log = (tau1Yplot.^fitcoeffy1log(1))*10^fitcoeffy1log(2);

fitcoeffy2log = polyfit(log10(fitTimeY(3):.01:fitTimeY(4)),logY(fitTimeY(3)*100:fitTimeY(4)*100),1);
liney2log = (tau2Yplot.^fitcoeffy2log(1))*10^fitcoeffy2log(2);

fitcoeffr1log = polyfit(log10(fitTimeR(1):.01:fitTimeR(2)),logR(fitTimeR(1)*100:fitTimeR(2)*100),1);
liner1log = (tau1Rplot.^fitcoeffr1log(1))*10^fitcoeffr1log(2);

figure;
plotgr;
axis([0 10 0 max(R)+10]);
fprintf('
Writing parameters to file: %s\n',[data1,'.dfp']);
fid = fopen([datapathc,data1,'.dfp'],'wt');
param = [critpointX*.01; X(critpointX); fitcoeffx1(1)/2; fitcoeffx1log(1)/2; critpointY*.01; Y(critpointY); fitcoeffy1(1)/2; fitcoeffy1log(1)/2; critpointR*.01; R(critpointR); fitcoeffr1(1)/2; fitcoeffr1log(1)/2];
fprintf(fid,'%f
',param);
close(fid);
set(h,'Pointer','arrow');
Third SDA Script:

function [fitTime, critpoint] = findtime(der2data)
fitTime(1) = 1; %First time is 0.0
[trash fitTime(2)] = max(der2data(1:100));
[trash critpoint] = min(der2data(1:250));

min1 = 0;
i = 900;
while min1 == 0,
if der2data(i-25) < der2data(i),
j = i;
while der2data(j-25) < der2data(j),
j = j - 1;
end
[trash min1] = min(der2data(j-25:j));
end
i = i - 1;
end
min1 = min1 + j - 25;
i = min1;
if der2data(i-25) > der2data(i),
j = i;
while der2data(j-25) > der2data(j),
j = j - 1;
end
[trash max1] = max(der2data(j-25:j));
max1 = max1 + j - 25;
else
[trash max1] = max(der2data(i-25:i));
max1 = max1 + i - 25;
end
if max1 < 700,
    max1 = 900;
end

fitTime(4) = max1;

maxlow = critpoint;
if der2data(maxlow + 25) <= der2data(maxlow),
    [trash maxlownew] = max(der2data(maxlow:maxlow + 25));
else
    while der2data(maxlow + 25) > der2data(maxlow),
        maxlow = maxlow + 1;
    end
    [trash maxlownew] = max(der2data(maxlow:maxlow + 25));
end
maxlow = maxlownew + maxlow;
fitTime(3) = maxlow;
fitTime = fitTime *.01;
Appendix B: Matlab Script for Detrended Fluctuation Analysis

This script performs a detrended fluctuation analysis on a nonstationary input signal to obtain an estimate for the scaling exponent. The first section is setup to allow input of a standard ASCII text or Matlab file. The script was based on previous work by Little et al. (2006) (76).

```
disp(' ')
disp(' Select file input method ');
disp('   1=external ASCII file ');
disp('   2=file preloaded into Matlab ');
file_choice = input('');
if(file_choice==1)
    [filename, pathname] = uigetfile('*.*');
    filename = fullfile(pathname, filename);
    fid = fopen(filename,'r');
    x = fscanf(fid,'%g %g',[2 inf]);
    X=X';
else
    THM = input(' Enter the matrix name: ');
end
%******************************************************************%
function [alpha, intervals, flucts] = fastdfa(x, varargin)
[xpts, ypts] = fastdfa_core(x, varargin{:});
% Sort the intervals, and produce a log-log straight line fit
datapts   = sortrows([xpts ypts],1);
intervals = datapts(:,1);
flucts    = datapts(:,2);
coeffs    = polyfit(log10(xpts), log10(ypts), 1);
alpha     = coeffs(1);
```
Appendix C: Matlab Code for Creating Stream Files.

This script builds a stream file to be read by an FEI Strata Dual-Beam system. The stream files define a square array of nano-holes with parameters defined by the user. Text following ‘%’ defines commenting.

```matlab
loop = 6;
npix=7.14;  %conversion factor for magnification of x5000 (i.e. spot
  %size of FIB at x5000 magnification is 7.14nm)
l= 3500;  %length of matrix pixels l=1000*NanoAr/npix
w= 3500;  %width of matrix pixels  w=1000*NanoAr/npix
NanoAr = l*npix/1000;  % array size in um (e.g: approx: 3502=25um,
  %4200=30um, 1401=10um, 7003=50um)
pmat = zeros(l,w,'uint16');
dia= 50;  % Hole diameter [nm]
per= 300;  % Resonance wavelength (periodicity) [nm]
dwt = 3000;  % Dwell time for milling [ns]
pskip= 2;  %Pixel skip factor
% Conversion from nanometers to pixels
diap=uint16(dia/npix);
radp=Diap/2;
perp=uint16(per/npix);
rad=uint16(dia/2);
arraymid=w/2;
%Builds Square Arrays
xx=0;
yy=0;
for y=(10+radp):perp:w
  for x=(10+radp):perp:w
    ppat(x,y)=dwt;
    if (x>xx)
      xx=x;
    end
  for yp=y:pskip:(y+radp)
```

76
for xp=x:pskip:(x+radp)
a=xp-x;
b=yp-y;
if (((yp-y)^2+(xp-x)^2)<=(radp^2))
    pmat(xp,yp)=dwt;
    pmat((x-a),yp)=dwt;
    pmat(xp,(y-b))=dwt;
    pmat((x-a),(y-b))=dwt;
end
end
end
end
count=0;
for x=1:w
    for y=1:w
        if (pmat(x,y)>0)
            count=count+1;
        end
    end
end
end
count
StrFile=strcat('ARRD',int2str(dia),'P',int2str(per),'DT',int2str(dwt),'A',int2str(NanoAr),'.STR');
fid=fopen(StrFile,'w+');
fprintf(fid,'s \n %u\n',loop);
fprintf(fid,'%u\n',count);
m=0;
for x=1:w
    for y=1:w
        if (pmat(x,y)>0)
            m=m+1;
            fprintf(fid,'%u\t %u\t %u\n',pmat(x,y),x,y);
            n(m,:)=pmat(x,y);
"
o(m,:) = x;
p(m,:) = y;
end
end
end
fclose(fid);

Arr = [n o p]; %prints array of data for stream file for cross checking of file
Appendix D: Examples of Centre of Pressure Plots from Force Plate data Collected from Young and Elderly Subjects.

The following are plots produce from the raw force plate data collected. This data was collected at 100 Hz over a 5 minute period. The plots illustrated show the forces exhibited in the mediolateral (x) and anterior posterior (y) directions of the subjects sway. The force exerted in the downward direction (i.e. gravitational acceleration multiplied by the subjects mass) was subtracted out – see section 2.1.2.2 for details. All units are in Newtons, and as can be see the forces exhibited in these directions are small and well within the range of an IPMC force sensor.
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