THE INTERACTION BETWEEN CARDIOVASCULAR AND POSTURE CONTROLS AND THE EFFECTS OF AGEING.

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ABSTRACT

Fainting during quiet standing is a serious problem in syncopal youth and prominent in elderly individuals. This thesis examined the possible interaction of the cardiovascular and posture control systems during quiet stance and the effect of ageing. We hypothesized that there exists a bidirectional interaction between blood pressure (BP) and posture changes through the activation of skeletal muscle pump via the calf muscles. In study one, the analysis methods for non-stationary signals were explored and the wavelet transform coherence method was validated for its applicability. In study two, cardiovascular variables, stabilogram data and posture muscle electromyography (EMG) data were compared to each other. The EMG and BP signals showed discrete regions of high coherence during quiet stance. In a third study, the effect of ageing on the relationship between these two systems was investigated. Age was found to have a statistically significant affect on the relationship between these two systems.

Keywords: Wavelet transform coherence; orthostasis; postural sway; Electromyography; skeletal muscle pump; blood pressure.

DEDICATION

To my parents, for always having faith in me and my actions

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GLOSSARY

SBP	Systolic Blood Pressure
DBP	Diastolic Blood Pressure
CO	Cardiac Output
EMG	Muscle Electromyography
COPx	Center of Pressure trajectory in Medio-lateral direction
COPy	Center of Pressure trajectory in Antero-posterior direction
WTC	Wavelet Transform Coherence
CWT	Continuous Wavelet Transform
SDA	Stabilogram Diffusion Analysis
LF	Low Frequency
VLF	Very Low Frequency
ULF	Ultra Low Frequency
EO	Eyes Open Condition
EC	Eyes Closed Condition

CHAPTER 1: GENERAL INTRODUCTION AND LITERATURE REVIEW

1.1 Introduction

Falls are a leading cause of injury related hospitalizations in Canada for all ages (CIHI 2004). Approximately 10% of falls result in serious injury including fractures, head injuries, and lacerations (Tinetti, Speechley et al 1988, Nevitt, Cummings et al 1991; Tinetti, Doucette et al 1995; Tinetti 2003.). The current financial cost of falls in Canada is estimated to be $3 billion annually. Furthermore, falls often result in permanent disability, which cost Canadians $1.1 billions annually (SAMRTRISK 1998).These statistics highlight the need for the development of improved programs to reduce the incidence and severity of falls.

Loss of balance and fainting are two of the symptoms associated with many conditions including concussion and syncope, falls and return from space. One in three people over 65 years old and one in two aged 85 years or over fall each year and about half of these falls involve bone fractures (Wilkins, 1999). Falls are a leading cause of morbidity and mortality in the elderly, and are the sixth leading cause of death. Often, these falls result from a loss of balance due to decreased postural control. The non-fatal results of falls include physical trauma, fear of walking, physical degeneration from immobility, and post-fall institutionalization. Prevention of falls would be a major step towards reducing the decline of mobility in the elderly.

Falls in the elderly are also more commonly associated with conditions such as arthritis, depression, orthostasis, the use of four or more prescription medications, and impairments in cognition, vision , balance, gait and muscle strength (Tinetti, Speechley et al 1988; Nevitt,Cummings et al 1989; Tinetti 1994; Tinetti 2003).

There is an age related decline in pressure and tactile sensitivity in the feet. This leads to a decline in the ability to sense and correct for changes in terrain, and contributes significantly to increased susceptibility to falls. Additionally, age related reductions in cardiovascular autonomic control and/or autonomic neuropathy can predispose one to hypotension during postural changes and may cause light-headedness and syncope. The present research was directed towards postural stability during quiet stance and cardiovascular regulation in response to exercising or working environmental conditions. Improved knowledge is required on how the cardiovascular system may be influenced by the postural control system, and vice versa, to better understand fall proneness under all conditions.

1.2 Postural and sensory-motor control system

Postural control involves the control of the body's position in space for the dual purpose of stability and orientation. Postural orientation is defined as the ability to maintain an appropriate relationship between body segments and between the body and the environment for a task (Horak and Macpherson, 1996). As postural control for stability requires both perception and action, it requires a complex interaction of musculoskeletal and neural systems.

Regardless of the type of perturbation, sensory information must be integrated from a variety of sources including somatosensory, visual and vestibular pathways to make appropriate balance corrections (Dichgans and Diener 1989, Rothwell 1994). The weighted contributions of various sensory inputs to balance responses is unclear, however, it has been suggested that the way sensory information is integrated for posture control must be flexible to accommodate changes in the sensory environment (McCollum 1996) and/or environmental circumstances (Woollacott and Shumway-cook 1990).

The sensorimotor system is a subcomponent of the comprehensive motor control system of the body. Humans are able to maintain a posture perfectly adapted to the environment with the help of a central, neural, multimodal system organized on the principle of the sensorimotor loop (Paillard, Jacques, 1976). This functional entity, called the fine postural system (FPS), is defined as the core neural system that provides balance control to the human body during static and dynamic performance (Paillard, Jacques, 1976). Key components of the FPS include the *visual-ocular system*, *vestibular system*, *foot mechanoreceptors, oculomotor* and *paravertebral muscles,* and *lower limb proprioception*. FPS coordinates postural neuromuscular tonus, visual-ocular motor controls, and visual-ocular-vestibular controls to naturally optimize body orientation and balance control during static and dynamic activities, and control posture, gait and related cognition (Gagey 1991).

The components that give rise to functional joint stability must be flexible and adaptable because the required levels of stiffness and forces production vary

between persons and among tasks. The process of maintaining functional joint stability, a particular concern for elderly health, is accomplished through a complementary relationship between static and dynamic components. Dynamic contributions arise from feedforward and feedback neuromotor control over skeletal muscles crossing a joint. Underlying 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 threats to standing balance. To maintain standing balance, the ability to detect postural disturbances and generate proper responses is required. These abilities have long been noted to deteriorate with increasing age, and could lead to imbalance and increased risk of falling (Blaszczyk et al., 1994; Tinetti et al., 1994; Mackey and Robinovitch, 2006; Hsiao-Wecksler and Robinovitch, 2007). Deterioration in the function of the somatosensory and motor systems also occurs with aging, and can be related to poor static standing balance (Hurley et al., 1998; Lord et al., 1991). However, in addition to older adults, altered reactive postural responses was also observed in healthy, young adults who were experimentally deprived of normal somatosensory function, and in patients with vestibular dysfunction or peripheral neuropathy. Limited contraction capacity in the gastrocnemius and smaller muscle response amplitudes have been observed in functionally unstable older adults compared with stable older adults when subjected to support surface perturbations (Lin et al., 2004). Thus, the control of reactive balance is also under

the influence of the function of the sensorimotor system-. Recent studies of postural stability in the elderly in relation to falls have revealed that mediolateral sway in balance control tests with narrow base stance may be an important tool in the identification of elderly fallers (Brauer et al., 2000; Melzer et al., 2004; Buatois et al 2006). Furthermore, subjects with multiple falls within 16 months of the balance tests had significantly more sway when visual inputs were removed (Buatois et al 2006).

1.3 Cardiovascular changes to orthostatic stresses

When upright, gravitational (orthostatic) stresses act on the cardiovascular system resulting in increases in blood pressure below heart level, and decreases in blood pressure above heart level. The difference between these pressure gradients increases as a person transitions from supine to semi-recumbent, seated, and upright positions. This results in up to 70% of human blood volume residing below the heart when upright due to blood pooling in the veins of the abdomen and lower limbs as a consequence of this gravitational gradient.

With mild orthostatic stress, blood redistribution from splanchnic organs (Rowell 1993) and peripheral musculature to systemic circulation occurs subsequent to a rise in sympathetic nerve activity via selective baroreceptor unloading (Baily et al. 1990, Laszlo et al. 1998, Mohanty et al. 1987). With more severe orthostatic stress, a further reduction in venous return occurs and is observed as decreased central venous pressure (Kimmerly and Shoemaker 2002). There is, thus, a further baroreflex-mediated increase in sympathetic nerve activity that induces a rise in heart rate and peripheral resistance in order

to compensate for reduced central venous pressure (Pawelczyk & Raven 1989). During severe orthostatic stress, over 800 ml of blood may be translocated from the central circulation into the lower limb and pelvic regions (Lundvall et al. 1993).

Arterial baroreceptors are crucial in the maintenance of cardiovascular control during orthostatic stress, and originate from mechanoreceptors in the carotid sinus, coronary arteries, and aortic arch (Jacobsen et al. 1993). Sympathetic discharge is transmitted to peripheral vessels via caudal ventro-lateral medulla and rostral ventro-lateral medulla (Robinson and Potter 1997). These receptors are known to be sensitive to changes in mean arterial pressure and pulse pressure on a beat-by-beat basis under conditions such as loss of blood or blood pooling during orthostatic stress. Reflex responses to hypotension involve two pathways: vagal withdrawal, increasing heart rate rapidly; and sympathetic activation of the heart and vascular system, initiating tachycardia and vasoconstriction over a slower time course (Rowell 1993). The goal of these two pathways is to maintain a constant mean arterial pressure. Upon the removal of orthostatic stress, opposite axis responses that decrease blood pressure are initiated, leading to increased vagal tone and decreased sympathetic drive (Rowell 1993). These baroreceptor reflexes are essential to control blood pressure during orthostatic challenges.

The frank starling mechanism established in the early 1900's states that greater the volume of blood entering the heart during diastole (end-diastolic volume), the greater the volume of blood ejected during the systolic contraction

(stroke volume). In other words, the cardiac output depends on the central venous pressure and any change in one affects the other. Under orthostatic stress, the blood redistribution to the peripheral vasculature leads to a reduced blood volume in central circulation, resulting in lower central venous pressure with reduced ventricular filling and hence a diminished cardiac output.

To compensate for reduced central venous pressure when upright, vasoconstriction is induced by enhanced muscle sympathetic nerve activity, which raises vascular resistance, and in conjunction with elevating heart rate, serves to maintain blood pressure. Activation of skeletal muscles in the lower limbs (skeletal muscle pump), such as through postural shifts, walking or running, also increases venous return by pumping blood collecting in the veins back to the heart. The maintenance of upright posture not only requires coordinated vestibular-neuromuscular control of postural muscles, but also cardiovascular reflexes to maintain blood pressure.

An upright position ordinarily produces subdiaphragmatic gravitational blood pooling, primarily within the venous system (Rowell 1993). As a result, venous return and cardiac output (CO) decrease in all upright subjects. In healthy subjects, baroreceptors sense the postural decrease through arterial and cardiopulmonary stretch, which maintains blood pressure through a compensatory increase in total peripheral resistance (TPR), elastic recoil of venous blood, and an increase in heart rate (HR). This is the normal reflex response to orthostasis (O' Leary et al., 2003).

1.4 Cardio-postural interactions and Syncope.

Until recently, cardiovascular and postural reflexes have been examined as independent control systems. There has not yet been an adequate in-depth analysis of the interactions between cardiovascular and postural control centers, particularly in relation to the importance of the skeletal muscle pump in the maintenance of venous return in conditions of impaired vascular control or increased venous pooling. Recent research has demonstrated that postural sway may be linked to the prevention of syncope in otherwise healthy subjects. Participants with poor orthostatic tolerance during a passive orthostatic stress test had greater leg movements during upright stance, possibly preventing them from fainting (Claydon and Hainsworth, 2005).

Simple postural syncope, such as upright vasovagal syncope, in which vasodilation causing hypotension and bradycardia is relieved by recumbency (Lewis 1932).

Increased (Jardine et al 1998), decreased, (Mitro et al 2006) or unchanged (Nowak et al 2007) sympathetic activity and TPR compared to non-fainting subjects have been reported during instances of postural fainting. Blood pressure often begins to fall well in advance of loss of consciousness, indicating inadequate compensatory sympathetic vasoconstriction, reduced venous return, or both. Ultimately, sympathetic withdrawal, vasodilation, and hypotension supervene, precipitating loss of consciousness (Lewis 1932).

There is a lack of extensive research in the direction of a possible interaction between the cardiovascular and posture control. There is a dearth of

scientifically structured studies with validated outcomes to explain the observations of increased postural sway in the non-fainting group and vice versa. The present thesis has attempted to bridge this gap and contribute to the understanding of the integrated control of human body during orthostatic challenges.

1.5 Objectives

The objectives of this thesis were as follows: 1) to investigate and validate a coherence analysis method for application to nonstationary cardiovascular, EMG and posturography signal sets; 2) to investigate the presence of a bidirectional relationship between the cardiovascular and postural controls in the young healthy population; and, 3) to investigate the effects of ageing on the relationship between the cardiovascular and postural controls.

The studies conducted to fulfil the objectives are presented independently in the three chapters.

1.6 Summary

Falls are a major problem in the elderly population, causing hospitalization, permanent disability, or even death. Early detection of fall proneness is necessary for a reduction in the number of people suffering from fall related outcomes. Research in the direction of biomechanics has been conducted over the past decades to understand falls and has helped explain the reasons for loss of postural stability. Cardiovascular researchers have also shown changes in regulatory control in the elderly. However, limited attention has

been focused on interactions between the cardiovascular and postural systems. As part of this thesis, the presence of interdependent relationship between these two systems was verified and parameters that characterise the relation were quantified in the young age group. Furthermore, the effects of ageing on this “cardio-postural interaction” were studied.

CHAPTER 2: VALIDATION OF WAVELET TRANSFORM COHERENCE METHOD FOR APPLICATION TO CARDIOVASCULAR AND POSTURAL CONTROL SYSTEM

2.1 Abstract

Continuous wavelet transform is efficient in the analysis of non-stationary and transient signals. This method has already been established as a valid tool to analyze blood pressure and electromyography signals. Time frequency maps obtained from the wavelet transform coherence estimator (WTC) can provide further insight into the time varying coherent behaviour of the two signals.

Based on models for physiological systems, each signal was simulated as band pass filtered white noise. Frequency-averaged coherence in three frequency bands between 0.1-0.005 Hz was estimated. The Morlet wavelet was used as the mother wavelet in the WTC; bias, standard deviation, and false negative rate were calculated over 1000 iterations of the simulated signals. The coherence threshold (T) was established using simulated, uncoupled signal pairs. Data were collected (n=1) during a 10 minute sit-to-quiet stance protocol with eyes closed. Coherence estimates were obtained for real signals and corresponding surrogate data sets.

Values of the statistical accuracy measures validated the efficacy of the estimator. Real signals showed significant coherence (>T) in all three frequency bands. Surrogate data analysis reiterated the presence of significant coherence.

The Wavelet Transform Coherence method objectively shows significant coherence between blood pressure and electromyography signals.

2.2 Introduction

Physiological time series are generated through a combination of complex systems. Often, these relational combinations are of interest to the scientific community.

The blood pressure (BP) signal has been extensively studied for the analysis of baroreflex sensitivity, respiratory sinus arrhythmia, and other heart conditions (Saul et al.1991). On the other hand, electromyography (EMG) signals from the lower leg muscles have been studied exclusively for exercise and posture assessments (Loram et al 2005). Our laboratory is interested in the activation of lower leg muscles in relation to blood pressure regulation via the skeletal muscle pump. Previous research in our lab (Blaber et al, 2008) has shown that mediolateral (ML) sway and BP are related to each other; however, the methodology to investigate the possible relationship between the cardiovascular and posture control systems has not yet been developed. When the classical coherence method was applied, significant coherence was observed in the frequency range of 0.03 – 0.07 Hz. The interrelation between the BP and EMG signals in a common frequency range is of interest to investigate their bidirectional modulation.

Both EMG and BP signals have been identified as non-linear and non-stationary in nature (Padmanabhan and Puthusserypady 2004; Voss et al 2009).

Furthermore, the interaction between the two systems is both time and pulse dependent, thus, requiring a more sophisticated approach from the classical time series of Fourier analysis.

The wavelet transform has been used in many forms to analyze physiological signals. Discrete Wavelet Transform (DWT) analysis and continuous wavelet transform (CWT) have been used for feature extraction analysis of EMG signals (Kumar et al 2003; Kilby and Hosseini 2004, Kilby et al 2006). The blood pressure signal has also been analyzed with wavelets. These include investigation of low frequency fluctuations in rats (Tsai et all 2004 neuroscience letters); peripheral blood circulation oscillations (Stefanovska 1999); and vasovagal syncope (Marrone et al 1999). The use of both DWT and CWT in cardiovascular signals has been reviewed by Addison (2005).

Rowell (1993) demonstrated that both the transition from supine to upright stance and induced postural sway have a greater effect on systolic blood pressure (SBP) than diastolic or mean arterial pressure. Therefore, for the present study, we chose to obtain the SBP time series from the BP waveform for WTC analysis.

The coherence function is a method that is used to assess the existence and strength of linear coupling between two signals in the frequency domain (Kay 1988). Classical coherence and correlation methods have been used to investigate the relationship between signals; however, signal stationarity is assumed. This stationarity assumption removes some dynamic and complex

characteristics required for continuous physiological adjustments to maintain homeostasis.

Wavelet transform coherence (WTC) is a known, though infrequently used, signal analysis tool for random-like, yet deterministic, signals created by complex mechanisms. It has been applied to weather in geophysics (Grinsted et al. 2004), turbulence in plasma physics (Van Milligen et al. 1997), and brain dynamics in physiology (Klein et al. 2006). Wavelet coherence is used to find transient correlations between signals. The most common use of wavelet coherence is to find correlated areas between signals that are uncorrelated a majority of the time. The wavelet transform coherence method provides information on the strength of coherence as a time frequency map. By doing so, related signal features can be obtained over specific frequency zones and time points. Desired resolution can be obtained simultaneously for each signal feature; higher temporal resolution at higher frequencies, and higher spatial resolution for lower frequencies.

The scope of present work was to utilize WTC as a tool for identifying the relationship between the primary signals of the two physiological systems. Bias and standard deviation measures were estimated in the frequency range of interest. Threshold and error rates were calculated in order to establish the baseline characteristics of the method for the signals under investigation. Finally, the WTC estimator was applied to real physiological signals to understand the behaviour of the two systems over time.

2.3 Methods

2.3.1 Mathematical overview

This section describes the method of wavelet analysis based on the Morlet wavelet utilized to find the WTC estimate. A more detailed review can be found in previous published work of Torrence and Compo (1998) and Grinsted et al (2004).

2.3.1.1 Wavelet transform coherence

The CWT is defined as the convolution of the scaled mother wavelet function with the analyzed function g(t)

$$W(s,\tau) = \int g(t)\psi_s(t-\tau)dt \tag{2.1}$$

The Morlet wavelet is the commonly used mother wavelet for the analysis of physiological signals. The function for the Morlet Wavelet (ψ_0), which is a plane wave modulated by a Gaussian window, is defined in equation 2.2.

$$\psi_0(\eta) = \pi^{-1/4} e^{i\omega_0\eta} e^{-\eta^2/2} \tag{2.2}$$

where ω_0 is the dimensionless frequency and η is the dimensionless time.

The discrete form of the CWT and the mother wavelet function for a sequence x_n, is shown in equations 2.3 & 2.4.

$$W_n(s) = \sum_{n'=0}^{N-1} X_{n'} \psi^* \left[\frac{(n'-n)\delta t}{s} \right] \tag{2.3}$$

$$\psi \left[\frac{(n'-n)\delta t}{s} \right] = \left(\frac{\delta t}{s} \right)^{1/2} \psi_0 \left[\frac{(n'-n)\delta t}{s} \right] \tag{2.4}$$

The Morlet coefficient, ω_0 defines the balance between the frequency and time resolution where ω_0 >6 is the minimum requirement as per the admissibility condition (Farge 1992). For our analysis and frequency resolution requirements, we tested the coherence estimator in 6< ω_0 <30 for its statistical acceptance.

The wavelet power density estimator of x_n is defined as:

$$W_n^{xx}(s) = W_n^x W_n^{x^*} \tag{2.5}$$

where W_n^* is the complex conjugate of the wavelet coefficient W_n.

For the mother wavelet, the scale to frequency transformation is defined through the Fourier wavelength (equation 2.6) (Torrence and compo 1998).

$$\lambda = \frac{1}{f} = \frac{4\pi s}{\omega_0 + \sqrt{2 + \omega_0^2}} \tag{2.6}$$

Based on previous work in our laboratory (Blaber et al 2008), analysis of the WTC was conducted for three frequency bands, namely, Low Frequency (LF) band, 0.1 – 0.05 Hz; Very Low Frequency (VLF) band, 0.05 – 0.01 Hz; Ultra Low Frequency (ULF) band, 0.01 – 0.005 Hz.

The squared wavelet coherence estimator is defined as the squared absolute value of the smoothed cross- wavelet spectrum, normalized by the smoothed power spectrum of the two signals.

$$\hat{C}_n^2(s) = \frac{\left|\left\langle W_n^{xy}(s) \cdot s^{-1} \right\rangle\right|^2}{\left\langle \left|W_n^{xx}(s)\right| \cdot s^{-1} \right\rangle \left\langle \left|W_n^{yy}(s)\right| \cdot s^{-1} \right\rangle} \tag{2.7}$$

The $\langle \bullet \rangle$ is the smoothing operator shown in equation (2.8) as defined by Torrence and Webster (1998), and equation (2.9) defines the cross wavelet transform of two time series x_n and y_n, where * denotes the complex conjugate.

$$\langle W \rangle = \left[(c_1 \omega_n(s) * e^{-n^2/2s^2})_n * c_2 \prod(\delta j_0 s) \right]_s \tag{2.8}$$

$$W_n^{xy} = W_n^x(s) W_n^{y^*}(s) \tag{2.9}$$

In equation (2.8), c_1 and c_2 are the normalization constants, and $\prod$ is the rectangular function (Grinsted et al 2004). For the Morlet wavelet, the value of δj_0 is equal to 0.6 for value of ω_0 equal to 6 (Torrence and Compo 1998).

2.3.1.2 Statistical estimations for the WTC method.

2.3.1.2.1 Theoretical coherence

The theoretical coherence estimation was based on the model of a single input, single output (SISO) of a linear time invariant (LTI) system (Pinna and Maestri 2001). The theoretical coherence between X(t) and Y(t) was given by:

$$\gamma^2(f) = \frac{1}{1 + G_{NN}(f)/G_{XX}(f)} \tag{2.10}$$

Where $G_{XX}(f)$ and $G_{NN}(f)$ were the spectral density functions of the input signal, X(t), and the noise that was added to get the output signal, Y(t), respectively. In order to apply the model to our time- frequency analysis, we used the wavelet power spectrum:

$$\gamma_n^2(s) = \frac{1}{1 + W_n^{NN}(s) / W_n^{XX}(S)} \quad (2.11)$$

Equation (2.11) shows that the theoretical coherence is effectively determined by the signal to noise ratio of the system. Therefore, the simplest way to obtain different levels of theoretical coherence would be to change the amount of variance of noise relative to the input signal, keeping all other parameters constant (Pinna and Maestri 2001).

2.3.1.2.2 Bias and standard deviation estimation

Statistical error of the WTC method with respect to the theoretical coherence needed to be calculated to validate the new method. Simulated signals were generated to closely resemble the real signal in the analysis. As our analyses involve two different types of signals, we independently simulated the EMG and BP (in this case, SBP) signals.

Similar to Bonato et al. (2001), we modelled the MyoElectric signal obtained from surface electrodes as a filtered Gaussian noise signal. A shaping filter was used as suggested by Stulen and Deluca (1981) with the following transfer function:

$$H(f) = \frac{k^2 fh^2 f^2}{(f^2 + f_l^2)(f^2 + f_h^2)^2} \quad (2.12)$$

$f_l =$ bandpass low cut off frequency

$f_h =$ bandpass high cut off frequency

$k = 1.699 / f_h$

The EMG signal was synthesized using the above transfer function with the low and high cut off frequency at 0.01 and 0.5 Hz, respectively.

The SBP signal was modelled as a bandpass filtered white noise signal in the band 0.01 – 0.5 Hz to simulate an arbitrary SBP signal. All signals were generated for 2400 data points at a sampling frequency of 10Hz to create a data length equal to 4 minutes. These signals formed the input, X(t), to the SISO model for a physiological system.

The principal input signals, X(t), were obtained as described above. In accordance with the SISO system model for most physiological systems, the output signal, Y(t), was obtained by the addition of Gaussian white noise with zero mean to X(t).

The bias and the standard deviation (SD) estimates for the modulus of the transfer function for the SISO system have been defined by Pinna and Maestri (2001) as:

$$bias(f) = (\frac{1}{N}\sum_{i=1}^{N}\left|\hat{H}_i(f)\right|) - \left|H(f)\right| = \overline{\left|\hat{H}(f)\right|} - \left|H(f)\right| \quad (2.13)$$

$$SD(f) \cong \sqrt{\frac{1}{N-1}\sum_{i=1}^{N}(\left|\hat{H}_i(f)\right| - \overline{\left|\hat{H}(f)\right|})^2} \quad (2.14)$$

These estimates were adapted to our time-frequency estimation using the wavelet transform coherence as the transfer function.

The bias and the standard deviation for our coherence estimate were obtained as

$$bias(k) \cong (\frac{1}{N}\sum{}^{N} C^2(k)) - k = \overline{C^2(k)} - k \quad (2.15)$$

$$s.d.(k) \cong \sqrt{\frac{1}{N-1}\sum{}^{N} (C^2(k) - \overline{C^2(k)})^2} \quad (2.16)$$

Where k was the theoretical coherence level (γ^2) from 0.05 to 0.95 in 0.1 steps, and $C^2(k)$ was the calculated WTC value associated with the theoretical coherence level equal to k. This procedure was repeated for the different wavelet coefficients ω_0 = 6,10,15,20 and 30.

Synthesized input/output signal pairs (N = 1000) were generated for EMG and SBP, respectively.

2.3.1.2.3 Threshold and error rates for the coherence estimator

The value of the coherence estimator for two completely uncoupled signals provides information about the significance threshold values for the particular coherence estimator. The threshold defines the value above which the coherence value will be considered significant and the two signals have linearly dependent behaviour at that time point.

To find the threshold for the WTC estimator, signals for the SBP and EMG were synthesized as defined above (section 2.3.1.2.2). Using the SISO system model, the output signals were obtained with added white noise, but the variance was kept at a level that gave an SNR<<1, which provided a theoretical band

coherence value close to zero. The input/output pairs were then created and checked for the threshold of WTC for different values of ω_0.

The coherence was estimated between each input/output pair and averaged over iterations to give a coherence time series; the empirical sampling distribution (frequency histogram) was computed for each frequency band. The threshold for zero coherence T(f) was set at the 100(1-α) percentile of the coherence sampling distribution, where α is the significance level of the statistical test kept at 95% confidence or 0.05 (Fisher and Van Belle 1993).

The error rates were found relative to the threshold values obtained from the previous step. The false negative rate (β) was defined as the number of WTC outputs lower than the threshold value when there was coupling between the input and output signals. The false negative rate was calculated for different variance values of the added noise in order to get theoretical coherence to range from 0.05 to 0.95 in steps of 0.1.

2.3.2 Data collection

Data collection and experimental protocol are explained in detail in Chapter 3, section (3.3.2).

Data were band pass filtered with cut off frequencies of 0.01 to 20 Hz to contain the data for the frequency range of interest. The R waves in the ECG waveform were detected, and the corresponding SBP time series was generated. All data were resampled at 10 Hz using interpolation.

2.3.3 Surrogate data analysis

2.3.3.1 Iso-spectral Data

An iso-spectral (IS) surrogate time series has the same power spectral distribution as the original time series (Theiler et al 1992). The IS series was generated by computing the Fourier transform of the original series. The frequency domain series was convoluted with a random phase $(e^{i\phi})$, with ϕ ranging within –π to π. This resultant series with random phase was subjected to inverse Fourier transform to obtain the time series. By randomizing the phase, any phase dependent relation between the two time series was removed. The IS surrogate series permitted the estimation of the number of observed interactions that were due to the linearly coherent structure of the data.

2.3.3.2 Iso-distribution Data

Iso-distribution (ID) time series has a random shuffle of the original time series, and, as a result, has a flat power spectrum (Blaber et al 1995). To obtain the ID series, random numbers were generated and assigned sequentially to each data point in the original series. Then, the random numbers were arranged in ascending order to get shuffled original series. The ID surrogates permitted the estimate of the degree to which the observed interactions may be due to completely random fluctuations (noise) in data.

The surrogate data sets for the real EMG signal were generated for each iteration for a total of N = 500 iterations, and a WTC estimate was found for each iteration.

2.4 Results

2.4.1 Bias and standard deviation

2.4.1.1 EMG data

The simulations were carried out as described in section 2.3.1.2 above. Bias and SD reduced with increasing levels of coherence. As evident from Figure2-1, ω_0 =6 showed minimum bias for all coherence levels. With increasing coherence, SD showed reduced values. There was a steep fall in the value for SD for low coherence values which changed to a steady reduction for higher coherence.

2.4.1.2 SBP data

There was a trend towards a reduction in bias with increasing coherence levels as seen in Figure 2-2. Minimum bias was observed for ω_0 =6 at all coherence levels. Values of SD increased for lower coherence levels, and reduced for higher coherence levels after reaching a peak in the middle region. For ω_0 =6, SD had a near steady value in the LF band.

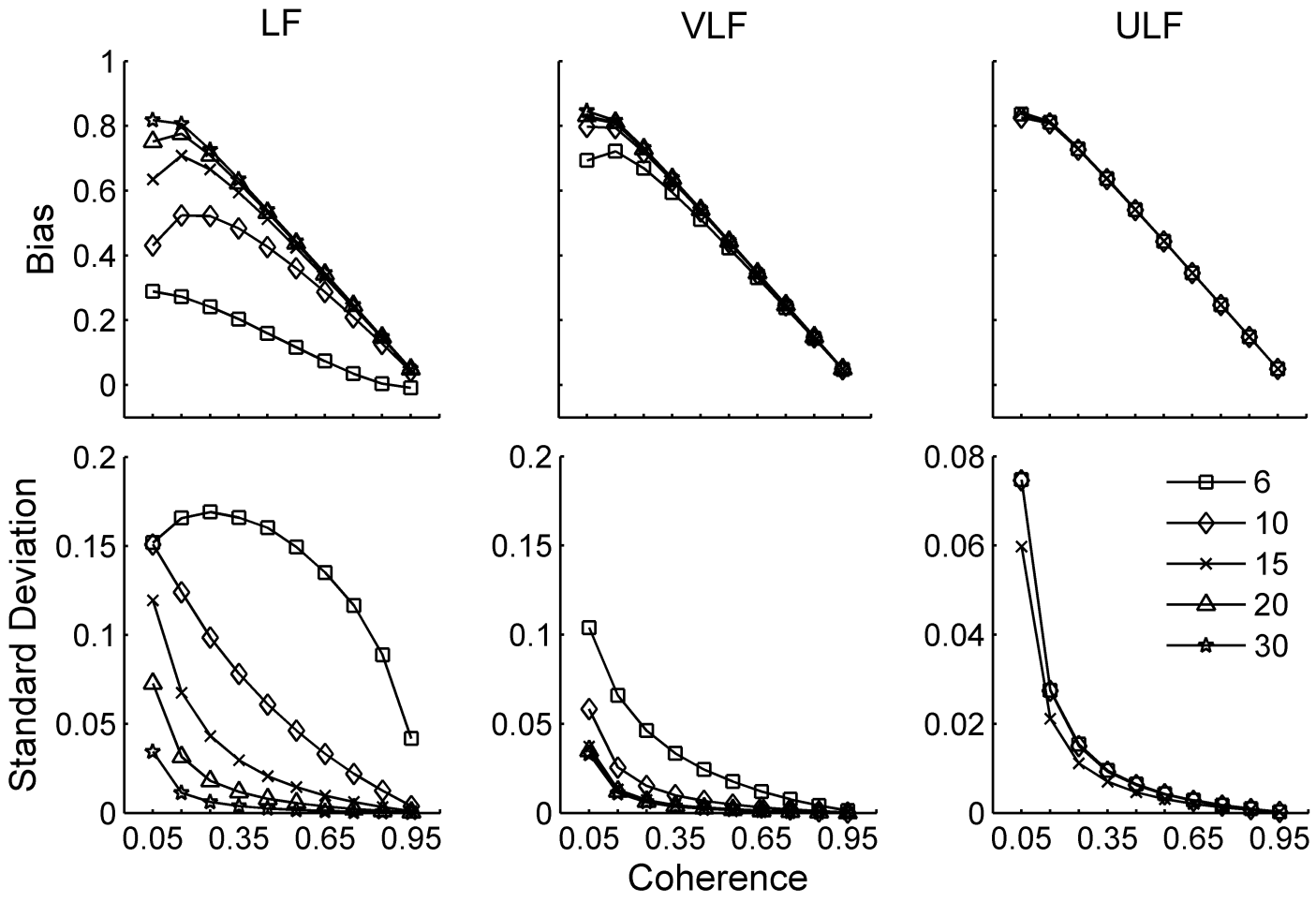

Figure 2-1 Coherence vs. Bias and Standard Deviation plots of the simulated EMG signals. ω_0 **=6 square;** ω_0 **=10 diamond;** ω_0 **=15 cross;** ω_0 **=20 triangle;** ω_0 **=30 star.**

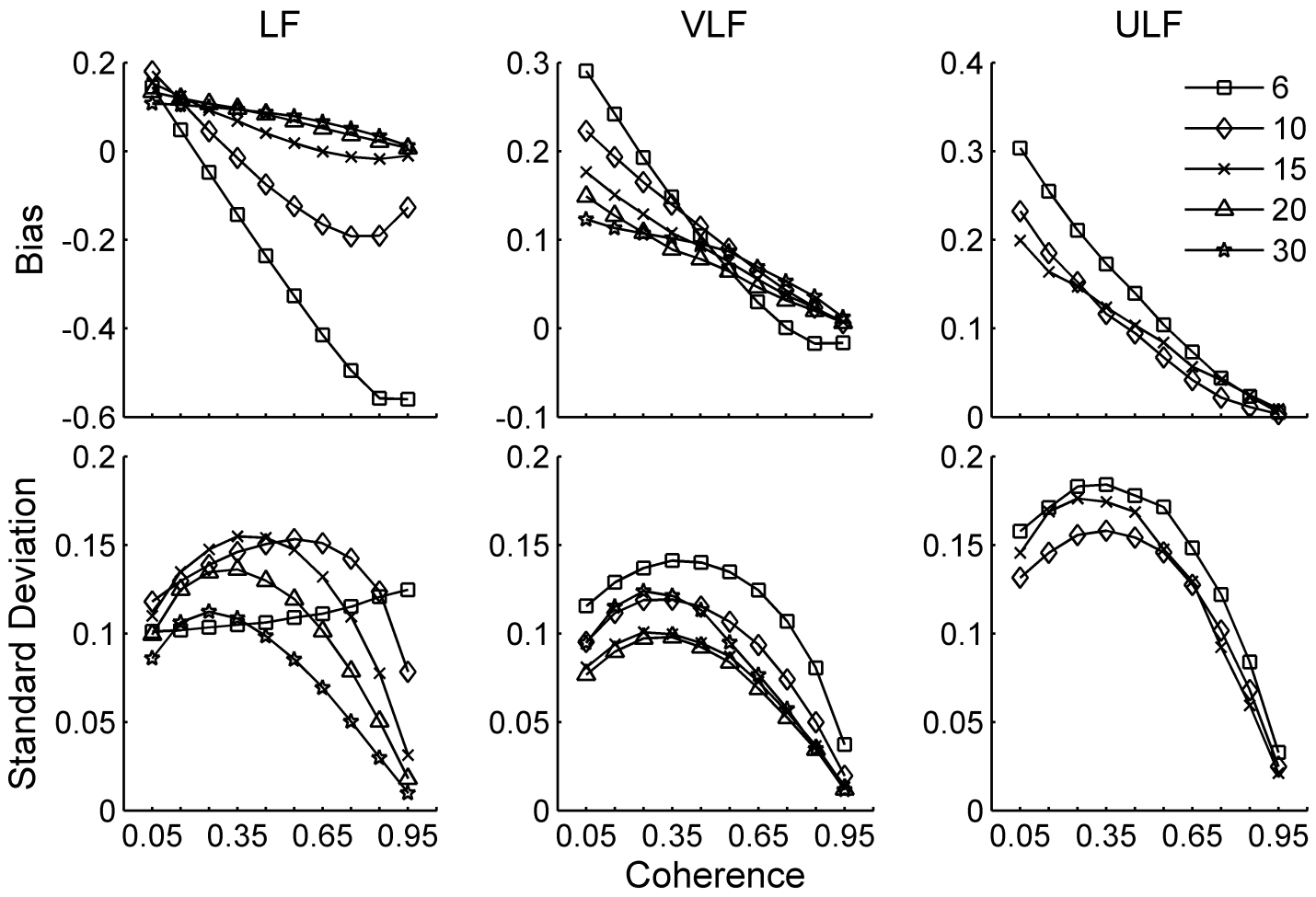

Figure 2-2 Coherence vs. Bias and Standard Deviation plots of the simulated BP signals. ω_0 **=6 square;** ω_0 **=10 diamond;** ω_0 **=15 cross;** ω_0 **=20 triangle;** ω_0 **=30 star.**

2.4.2 Error estimates

The threshold of coherence as defined in section 2.3.1.2.3 was calculated for simulated signals. After a thorough analysis of the threshold values for different values of ω_0, we chose threshold values of 0.1894 (LF), 0.3162 (VLF), and 0.3141 (ULF) corresponding to a value of ω_0=6.

The calculations for the bias and SD show that there is no ULF component for the calculation with ω_0 values of 20 and 30. Hence, we evaluated the error estimates for ω_0 values of 6, 10, and 15.

The percentage error for simulated BP signals was high for very low coherence (Figure 2-3), then returned to zero for higher levels of coherence. High error values for low theoretical coherence levels are as per the expectation due to the threshold values for coherence estimator.

The estimator showed no error for the simulated EMG signals (Figure 2-4) for all coherence levels in all frequency bands. The very high error value for low coherence level in the LF band was due to the low coherence threshold value. The presence of very low (≅ zero) error for the EMG signal was expected due to the signal characteristics.

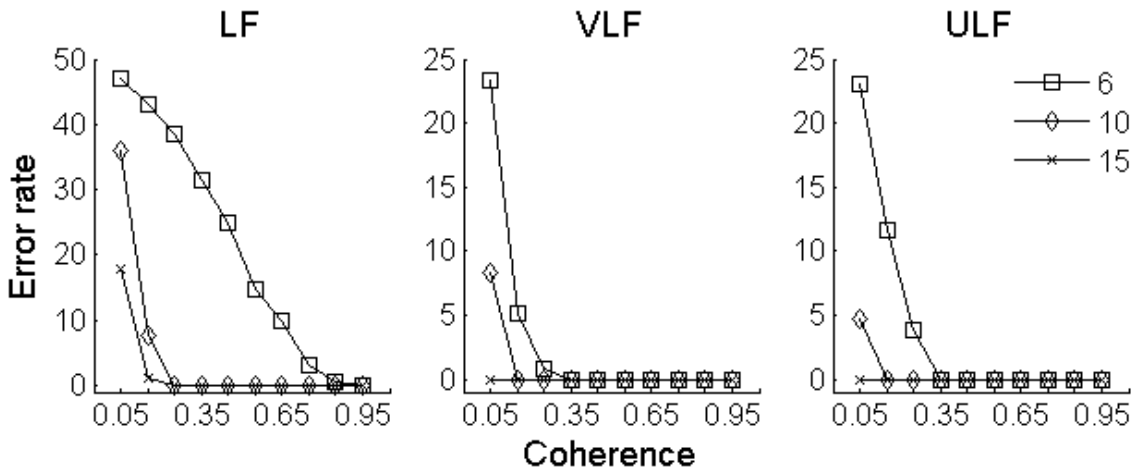

Figure 2-3 Coherence vs. false negative rate plot of the simulated BP signals. ω_0 **=6 square;** ω_0 **=10 diamond;** ω_0 **=15 cross.**

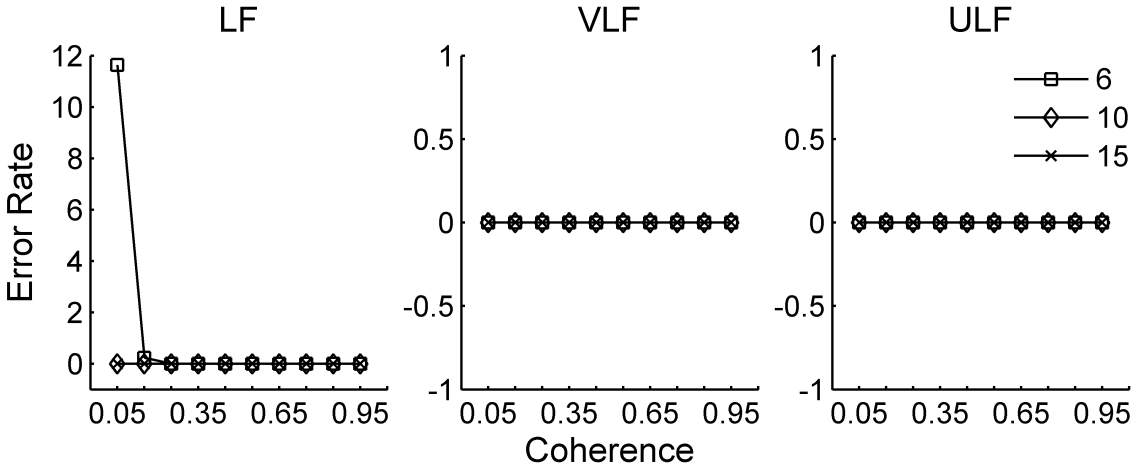

Figure 2-4 Coherence vs. false negative rate plot of the simulated EMG signals. ω_0 **=6 square;** ω_0 **=10 diamond;** ω_0 **=15 cross.**

2.4.3 Application to surrogate data

Data for the IS surrogates (Figure 2-5) show greater than threshold coherence in the LF and ULF bands. Similar behaviour (Figure 2-6) was shown by the WTC estimates for the ID surrogates. The coherence plot for the ID surrogates have much lower values than the IS surrogates, which implies that the occurrence of coherence is primarily due to signal characteristics than the presence of random noise in the signal.

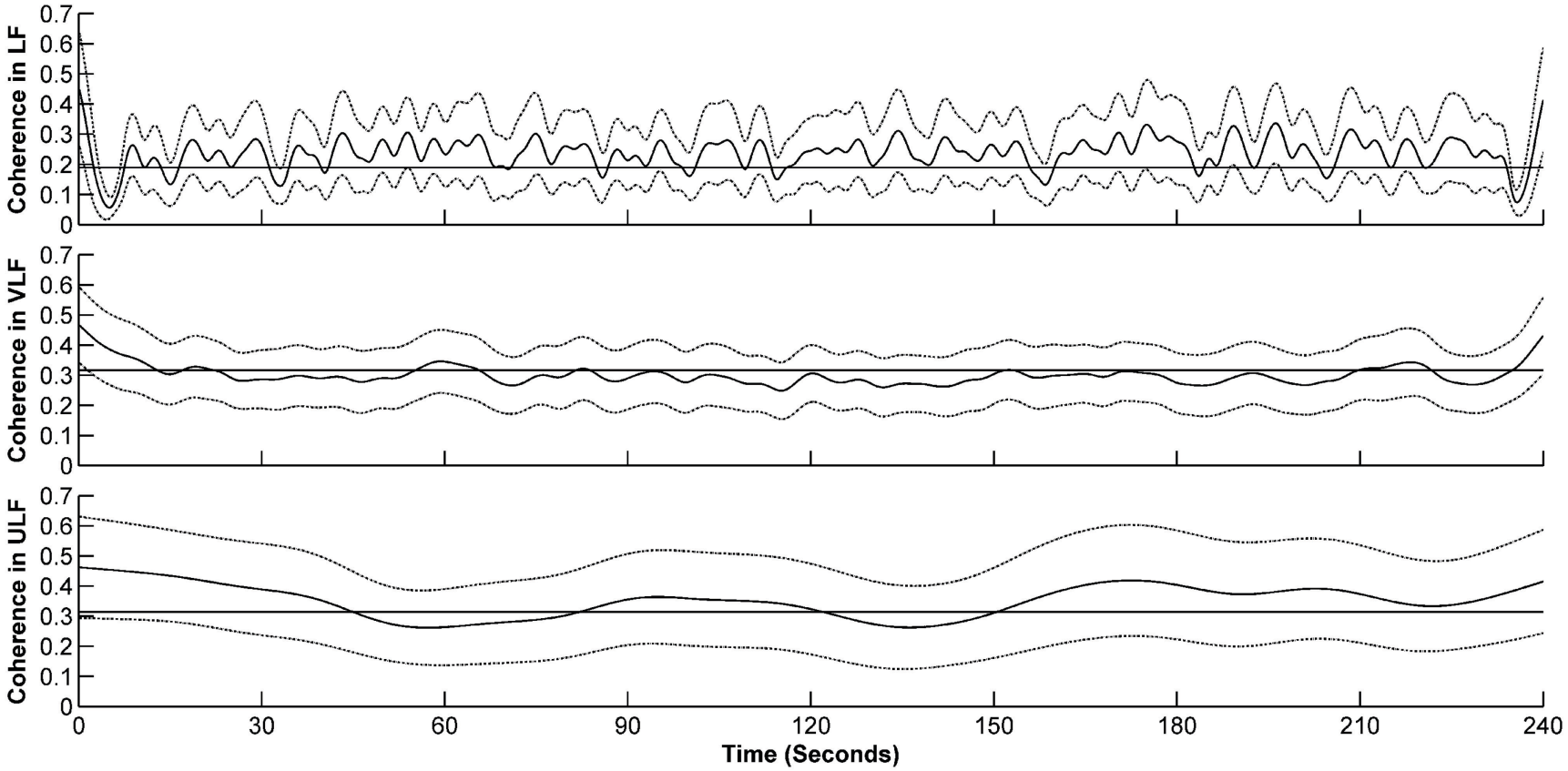

Figure 2-5 WTC output for the iso-spectral surrogate data in the three frequency bands (n=1). LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz).The straight lines indicate the significance level for each frequency band; the solid line is the mean; and dash-dot line is the ± standard deviation of the WTC estimate over the 500 iterations.

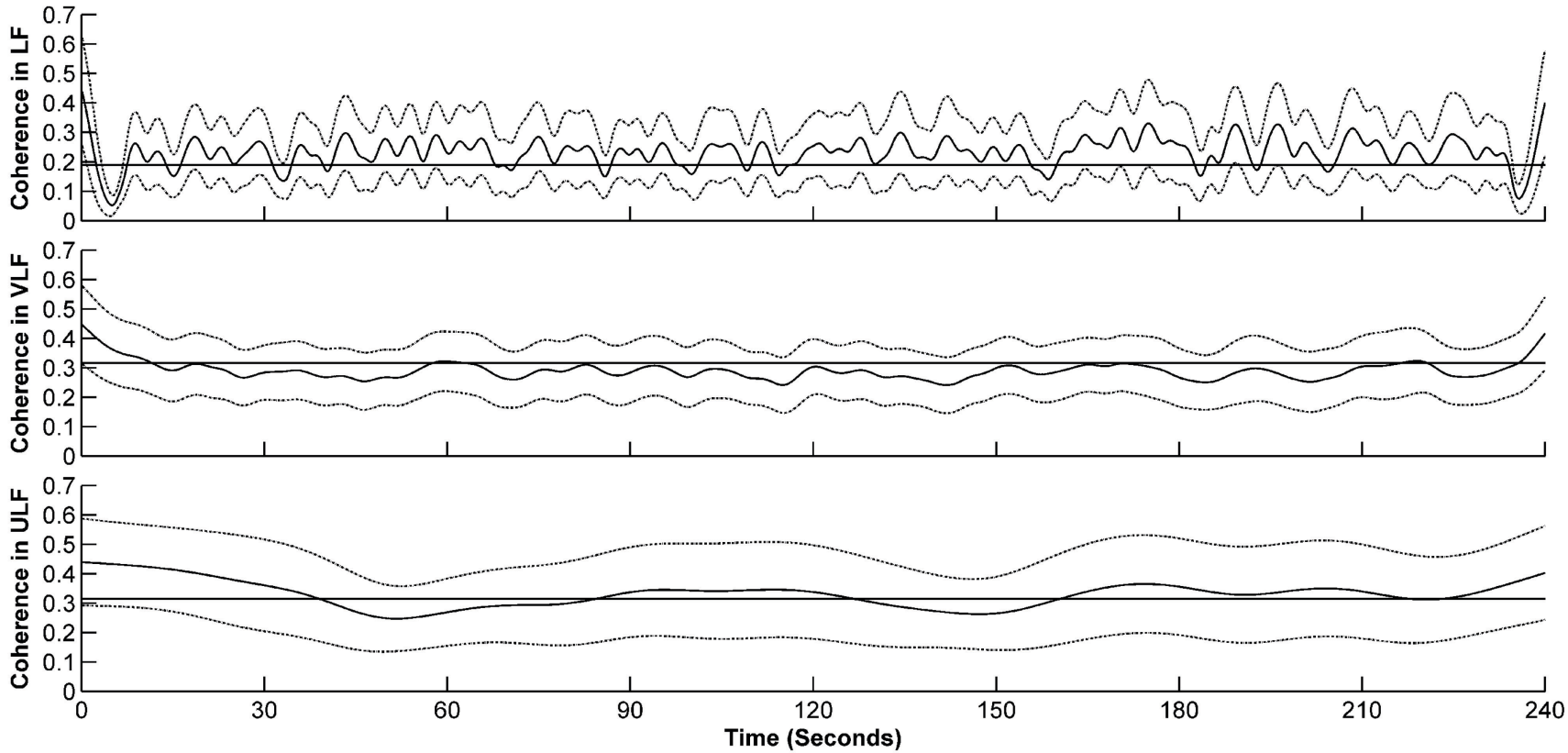

Figure 2-6 WTC output for the iso-distribution surrogate data in the three frequency bands (n=1). LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). The straight lines indicate the significance level for each frequency band; the solid line is the mean; and dash-dot line is the ± standard deviation of the WTC estimate over the 500 iterations.

2.4.4 Application to real data.

Comparison of EMG to BP signals showed significant, higher coherence in the LF and VLF bands (Figure 2-7). It is interesting to note that the coherence time series is above significance levels for at least one of the three frequency bands at any time point in the whole duration under analysis. This indicates there are changes in the spectral characteristics of the dependence between the two signals. The coherence estimator showed significant values for 54.8% (LF), 40.4% (VLF), and 42.3% (ULF) of the whole time duration under consideration.

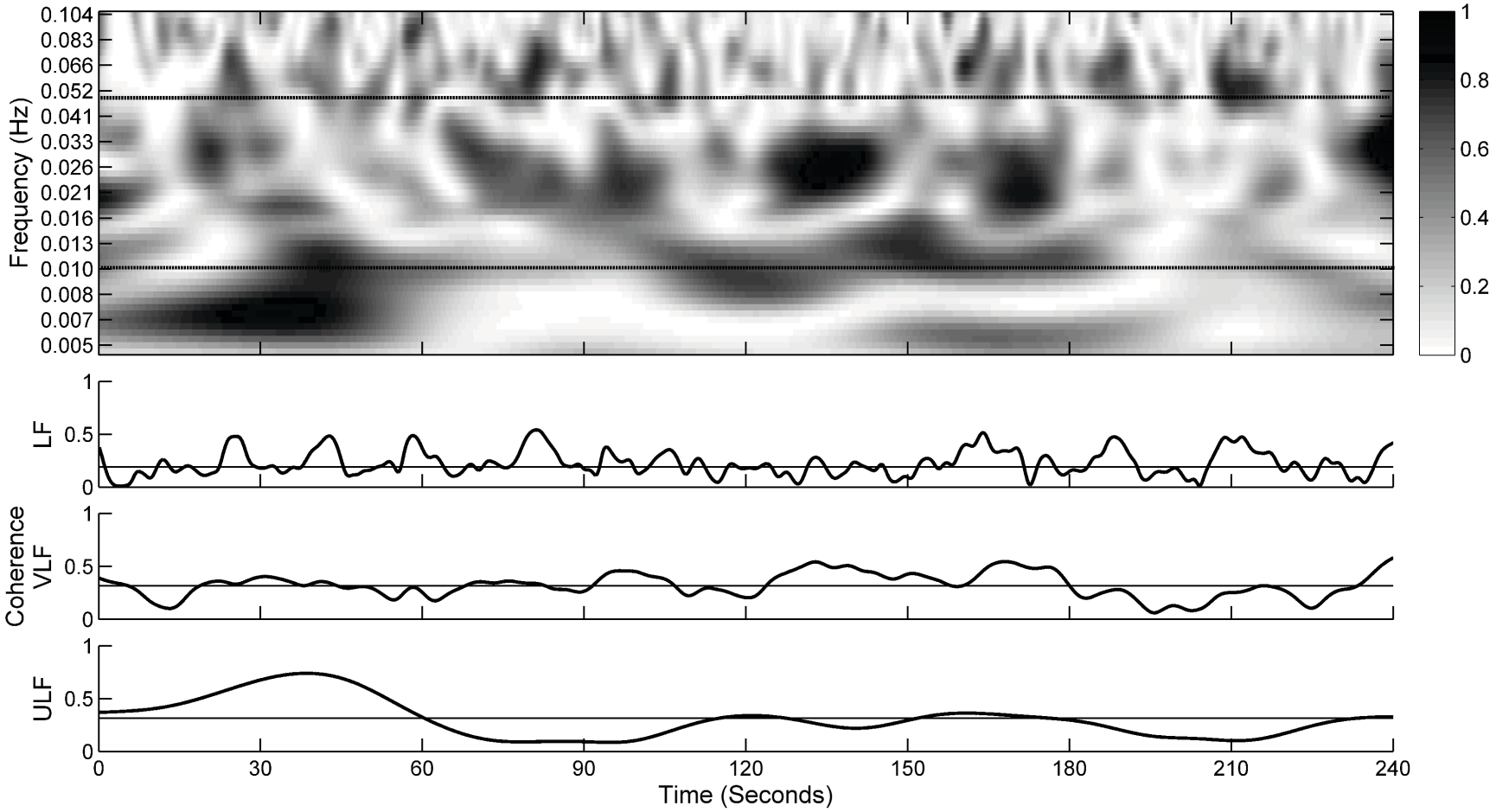

Figure 2-7 Real data analyzed using WTC method (n=1). The top figure is the time -frequency map of coherence between EMG and BP obtained from the WTC estimator. The bottom three plots represent band coherence obtained from averaging over corresponding frequency bands. LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz).The solid line is the estimate between EMG and BP; and solid straight line indicates the significance level.

2.5 Conclusion

The present study investigated the applicability of the wavelet transform coherence method to detect the existence of a relationship between changes in blood pressure and muscle activation in the lower legs during quiet stance in a young, healthy individual. Time frequency coherence maps were obtained for both simulated and human signals to quantify and validate the estimator. The performance was evaluated in the LF, VLF, and ULF frequency bands. These bands were chosen in accordance with the peaks in cross spectral analysis of the blood pressure and EMG signals (Blaber et al 2009). We chose to produce two representations of the coherence estimate (a) time-frequency maps of coherence estimate and (b) average band coherence time series for the signal combinations.

This investigation showed the WTC estimator to have low values of bias, standard deviation, and false negative rate for the simulated EMG and BP signal data sets. Analysis of real data with the WTC estimator indicated the presence of a relationship between the two signals. Both iso-spectral and iso-distribution surrogate data sets reiterated the presence of a relationship between the two signals. The plots in Figures 2-5, 2-6, and 2-7 suggest that, there is a strong dependence between the two signals in the LF and VLF region, which further suggests the occurrence of one to six events of coupled activity of the cardiovascular and postural controls in a one-minute duration. Further analysis is needed to characterize the physiological dependence between blood pressure and lower leg muscle electromyographic activity.

CHAPTER 3: INTERRELATIONSHIP BETWEEN THE CARDIOVASCULAR AND POSTURAL CONTROLS: YOUNG PARTICIPANTS.

3.1 Abstract

The cardiovascular system has been observed to respond to changes in human posture and the environment. Common observations of cardiovascular deficits in frequent fallers have been noted by many researches. The present study was designed to demonstrate the existence of a bidirectional relationship between the cardiovascular and postural systems through signal processing methods and statistical comparison techniques. Wavelet methods developed and verified in chapter 2 were applied to population data and statistical validation was performed. Existence of the bidirectional relationship was confirmed through high levels of coherence, phase locking periods with phase lead, and lag characteristics. The existence of a relationship between cardiovascular and postural control was observed in young individuals.

3.2 Introduction

The cardiovascular system has been shown to respond drastically to changes in orthostatic conditions in humans (Rowell 1993).

Activation of skeletal muscle activity in the lower limbs (skeletal muscle pump), such as through postural shifts, walking or running, also increases venous return by pumping blood collecting in the veins back to the heart (Rowell

1993). The maintenance of upright posture not only requires coordinated neuromuscular control of postural muscles, but also cardiovascular reflexes to maintain blood pressure.

There is a bidirectional interaction between blood pressure changes and posture changes through the contraction of the tricep surae muscle group . Since blood accumulates in the lower legs during upright posture due to the effects of gravity and venous compliance, the contraction of tricep surae group associated with posture control will force this blood back to the heart through what is termed the “skeletal muscle pump.”

The linkage between cardiovascular control and posture control via activation of the skeletal muscle pump is of interest in the present evaluation. Time domain analysis and Fourier transform based spectral analysis techniques were tested in pilot studies (Appendix A). These methods provided insights into the further development of systems analysis. The Fourier methods originally used were found to be inappropriate for the signals under consideration as they assume signal stationarity over time. As EMG signals by nature are non stationary, alternate methods workable for non-stationary signals were required to carry out further analysis.

Wavelet methods are used to analyze non-stationary EMG signals (Kumar et al 2003, Kilby et al 2006) and BP signals (Stefanovska 1999). Addison (2005) has reviewed the use of both discrete and continuous wavelet transform in cardiovascular research. The WTC method is a candidate in physiology

research. It has been applied in geophysics (Grinsted et al 2004), neuroscience methods (Lachaux et al 2002) and cardiovascular research (Keissar et al 2009).

The WTC method was tested for the signals under consideration (Chapter 2). The validation statistics, threshold and error rates were established and applied to analyze participant data in order to understand the relationship between EMG and blood pressure signals.

This analysis was utilized to determine the relationship between the EMG to center of pressure (COP), systolic and diastolic blood pressure (SBP, DBP) and cardiac output (CO); all of these have been shown to vary with changes in orthostatic condition (Rowell 1993, Winter et al 1996).

This chapter focuses on the relationship between the cardiovascular and posture control systems through their respective signals.

3.3 Methods

3.3.1 Data analysis methods

3.3.1.1 Wavelet transform coherence.

The WTC estimates for the signals were calculated (section 2.3.1) for the relationship of EMG to the center of pressure (COP), CO, diastolic blood pressure (DBP) and SBP. The same value of the wavelet coefficient (ω_0=6) was used throughout the analysis. Averaging over scales provided information about the band-averaged coherence estimate. The band coherence was calculated in the three frequency ranges, namely LF (0.05 – 0.1 Hz), VLF (0.001-0.05 Hz) and ULF (0.005 – 0.01 Hz).

3.3.1.2 Wavelet phase estimation

The phase dependence between the EMG and SBP signals was determined by wavelet phase estimation. The phase difference was obtained from the cross wavelet transform as defined below in equation (3.1)

$$W_n^{xy} = W_n^x(s)W_n^{y^*}(s) \tag{3.1}$$

The cross wavelet power estimation was not used as a direct comparison measure due to the lack of normalization (Maraun and Kurths 2004). The estimator may give spurious peaks which may not indicate a common power and be solely due to the dominant effect of one of the signals.

The phase information was calculated from the estimator for cross wavelet power (equation 3.1) and is defined as equation (3.2)

$$\theta_{xy} = \tan^{-1}[\Im\{W_n^{xy}(s)\}/\Re\{W_n^{xy}(s)\}] \tag{3.2}$$

This provided a time frequency map of the change in phase angle between the two signals under consideration. Averaging over scales corresponding to the frequency bands (LF; VLF; ULF) yielded the variation of phase in the corresponding frequency region over time.

All phase differences were defined in relation to the EMG signal (EMG phase – signal phase). A positive value of phase difference implied a lag of the signal from EMG, and vice versa.

3.3.1.3 Phase locking

The two signals were considered phase locked when the phase difference between them remained constant. Under the condition of phase lock, the two signals were considered to be interdependent. In order to find the existence of phase lock between the signals the phase signal was differentiated to obtain the rate of change of the phase angle. A rate of change of phase angle equal to zero corresponded to a constant value of the phase angle. The statistical equivalence to zero was found using a student's T-test. The 95% confidence interval obtained from the test provided a band within which the value was considered to be statistically, not different from zero and was used to find the presence of phase locking between the signals. This criterion was followed for all the signal combinations for each participant.

3.3.1.4 Stabilogram diffusion analysis.

Stochastic systems approaches have become a more popular method to analyze the fractal nature of COP data and other physiological time series such as heart rate. One of the first, and most commonly used techniques in force plate analysis, is stabilogram diffusion analysis (Collins and De Luca, 1993). 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).

The diffusion coefficient is an average measure of stochastic activity of a random walker, i.e. it is directly related to its jump frequency and/or amplitude.

The value of the diffusion coefficient (equation 4) provides the indication for the amount of randomness in a signal over the entire duration under consideration. A higher value of the diffusion coefficient implies a greater amount of randomness in the signal.

Modelling the COP trajectory as Einstein's classical law for Brownian motion, the relation between the displacement and time can be obtained as defined in equation (3.3). The value for the scaling exponent, *H,* is equal to 1/2 for classical Brownian motion and is bound as 0<H<1.

$$\left\langle \Delta x^2 \right\rangle \approx \Delta t^{2H} \tag{3.3}$$

In cases of fractional Brownian motion, past events are correlated with future events with a correlation function defined as (equation 3.4) (Feder 1988):

$$C = 2(2^{2H-1} - 1) \tag{3.4}$$

For H > 1/2, the stochastic process is positively correlated, i.e. C > 0. In this case, the fractional Brownian particle tends to follow the trend as in the event in the preceding moment. In case H < 1/2, the past and future increments are negatively correlated. Thus, the future event tends to follow the opposite trend than its preceding time point.

$$\left\langle \Delta x^2 \right\rangle \approx 2D\Delta t \tag{3.5}$$

$$\left\langle \Delta x^2 \right\rangle_{\Delta t} \approx \frac{\sum_{i-1}^{N-m} (\Delta x_i)^2}{N-m} \tag{3.6}$$

Equations (3.3) and (3.5) represent the mean square planar displacement between all pairs of the entire COP time series, separated by the time interval, Δt. Plotting Δx as a function of Δt (equation 3.6) yields a 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 scaling exponent (H). Analysis of a COP time series in this manner defines higher and lower levels of stochastic activity which can be interpreted as strategies of open loop and closed loop control, respectively (Collins and De Luca, 1995; Chiari et al, 2000).

In the present situation, the signals of the COP for the entire duration of 4 minutes were subject to the SDA analysis, and the corresponding outputs were compared through statistical analysis.

3.3.2 Data acquisition

The protocol was approved as minimal risk by Simon Fraser University's research ethics board. Written informed consent was obtained from each participant prior to the experiment.

3.3.2.1 Signals acquired

Electromyography: Bilateral lower leg electromyography was performed for four leg muscles: tibialis anterior, medial gastrocnemius, lateral gastrocnemius, and medial soleus. These were chosen in accordance with the observations by Joseph *et al.* (1955) for production of electric potentials during standing at ease.

Transdermal differential recording of signals was performed using an 8-channel EMG system, (*Myosystem 1200,* Noraxon Inc., Arizona, USA). For signal transduction, Ag/AgCl dual electrodes (2 cm inter-electrode distance) were used at the muscle sites, and a single Ag/AgCl electrode was placed at the right lateral malleolus as a reference electrode.

The sites for electrode placement were chosen in accordance with the recommendations for placement of electrodes from the SENIAM project (Hermens et al 1999).

Electrocardiography: ECG signals were acquired (LifePak 8, Medtronic Inc, Minnesota, USA) using the Lead II configuration of ECG electrode placement.

Blood Pressure: Blood pressure signals were acquired by photoplethysmography using a finger cuff electrode (Finapres, *Ohmeda 2300* Ohmeda, Ohio, USA).

The Finapres system functions on a principle of dynamic unloading of the arteries in the finger. The size of the finger arteries is gauged with an infrared transmission plethysmograph mounted inside an inflatable cuff. If the size of the finger arteries increase due to a blood pressure increase, the air pressure in the cuff also increases just enough to keep the arterial size constant. The pressure in the cuff is measured with an electric pressure gauge, and recorded after being processed through computer algorithms. (Langewouters et al 1998)

Postural Sway: The postural sway data, in terms of the coordinates of the COP of the body, were calculated from the force and moment data collected with a force platform (Accusway, Advanced medical technologies Inc, MA, USA).

The feet were placed in a parallel foot configuration with a distance of 5 cm between the first toe and heel of each foot.

3.3.2.2 Data collection

The data were acquired using a National Instruments data acquisition platform. This comprised of a 32-analog input channel BNC distribution box (BNC-2110), 68-pin shielded cables (SHC68-68-EPM), 32-analog input channel DAQ card (PCI-6229), Intel Pentium based personal computer using Windows operating system, and Labview 8.2 software platform, all manufactured by National Instruments Inc., TX, USA. A custom virtual instrument (VI) was designed using the in-built libraries. The system was configured to acquire data at 1000 Hz sampling rate and 16-bit analog-to-digital conversion. The VI provided output in the form of Windows text files. All variables recorded were stored as separate columns with header information for the participant details. All devices were connected to the data acquisition system through shielded BNC cables.

The experiment was conducted in a sensory input reduced environment within an enclosed space of black drapes to remove all random visual stimuli. Apart from equipment sounds, there was no noise in the room.

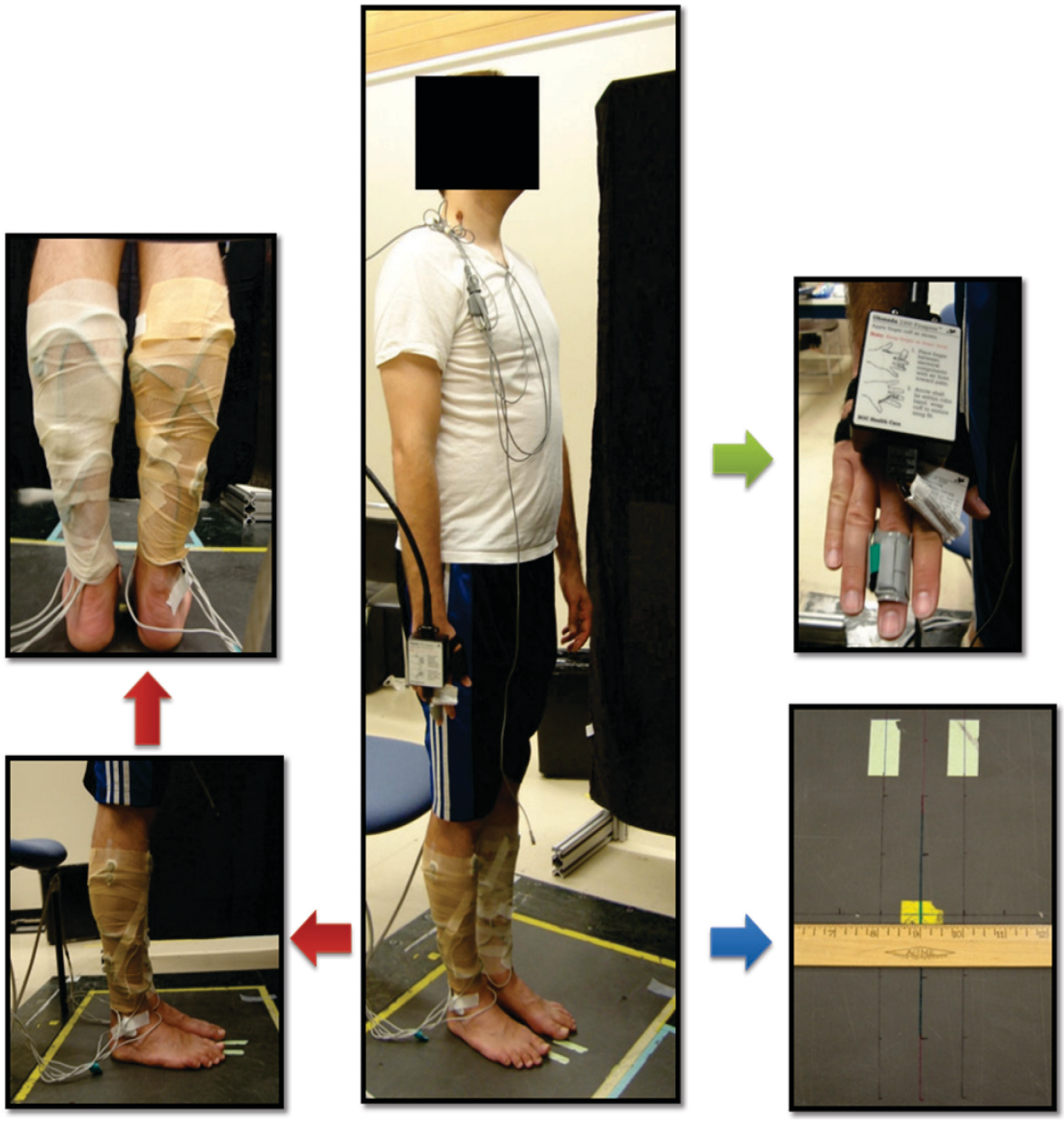

Figure 3-1 Quiet stance positioning and instrumentation: *(Centre)* Profile view. *(Left)* EMG electrode placement as seen from the rear and side view. *(Right, top)* BP finger cuff electrode attached to the middle finger of the participant. *(Right, bottom)* Heel positioning on force plate. A total distance of 5 cm between the toes of both feet.

3.3.2.3 Data preprocessing

Data pre-processing and analysis were performed using MATLAB 2009b (Mathworks Inc, MA, USA). All recorded data were converted into corresponding units from digital data in the number of bits format. Converted data were filtered with a Butterworth lowpass filter of fourth order with a cut-off frequency at 20 Hz for the frequency range of interest. The R-waves in the ECG waveform were detected, and the corresponding time mapped SBP time series was generated. All data were re-sampled at 10 Hz using interpolation before further analyses were performed.

3.3.2.4 Experimental protocol

Initial Setup: The participants were made to change into loose, comfortable clothing (shorts and a t-shirt) with empty pockets and no shoes. Measurements for height, weight and orthostatic correction for blood pressure were conducted prior to setup. Participants were set up for data acquisition prior to experimentation. The electrodes corresponding to all measurements mentioned in section (3.3.2.1) were affixed to the participant. After all the electrodes were placed, the participant was asked to sit quietly with a straight back and arms relaxed by the sides to verify the signal authenticity. They were instructed to not make any voluntary movements during the test and ask for assistance in the presence of any discomfort or uneasiness in which case, data collection would immediately cease. After visual verification of the signal quality, we proceeded to the sit-to-stand test.

Sit to Stand test: The participants were required to be seated for 5 minutes, after which they were asked to stand (assistance was provided during the transfer from sit to stand) for 5 minutes with eyes open. They were instructed to make a passive transition from the seated to upright stance phase without altering their foot position. During the test duration, they were required to maintain eye-level gaze. The same 10-minute procedure was repeated with eyes closed with an imaginary eye-level gaze in the same position as with eyes open.

3.3.2.5 Participant pool

Data were collected from young individuals between the age of 20 to 27 years. The participant pool comprised of 8 females and 9 males, yielding a total of 17 volunteers. All participants were screened for any cardiovascular disease or postural complications through verbal confirmation. All participants were required to refrain from exercise and caffeine for 24 hours prior to the experiment.

3.4 Results

3.4.1 Data pre-processing

Data for 2 female and 2 male participants were discarded due to incorrect recordings for the blood pressure waveform. All excluded participants had small fingers and the most probable explanation for the error is that, blood pressure finger cuffs were only available for medium, large, and extra large sizes. The incorrect finger cuff size could result in random fluctuations in the waveform or yield extremely high/low BP values.

The comparison for the eyes open versus eyes closed condition was not conducted on any of the 13 participants due to a problem in the recording of the BP channel in the eyes open condition. The COP data were collected with consistency and could be used for all participants in the SDA analysis.

Table 3-1 Anthropometric data for the young participant group

S. No.	Participant No.	Gender	Age (years)	Height (cm.)	Weight (kg.)
1	2	M	23	178	87
2	4	M	26	176.5	69.5
3	5	F	22	157.5	40
4	7	F	24	167.5	63.5
5	8	M	27	175.5	63
6	9	F	24	165.5	59
7	10	M	23	194	90
8	11	M	24	180	76
9	12	M	24	178.5	60
10	14	F	27	169	53
11	15	F	24	160	55
12	16	F	27	169	61.5
13	17	M	28	179	89.7

3.4.2 Wavelet transform coherence

The WTC outputs, in terms of the band coherence time series and the phase locking estimate, were obtained (sections 2.3.1 & 3.3.1). The outputs obtained were plotted with their corresponding signal sets and are explained in figures (3.2) and (3.3). Figure (3.2) shows the increase in coherence level to be greater than the significance level where the signals have a similar behaviour.

Statistical comparisons were applied to average phase angles (Table 3-2) to investigate the causal relationship between the variables and tested for an

association with the eyes condition in the three frequency bands. All phase differences were referenced against the EMG signal phase (EMG phase – signal phase). A positive value of phase difference implied a lag of the signal from EMG and vice versa. All phase angle values are in degrees.

The SBP lag was lower in the eyes closed condition (ec: 32.4± 3.1; eo: 40.9 ± 3.1; $p<0.05$) (Table 3-2). Lead for SBP was low in the LF and VLF bands, but it was significantly higher in the ULF band (LF: -39.3 ± 3.1; VLF: -35.8 ± 3.1; ULF: -50.7±3.1; $p<0.05$). Average phase for SBP and DBP for the overall phase lock period did not show any association with the eyes condition and frequency. Lead for DBP was significantly lower in VLF (-32.4 ± 3.1) compared to ULF (-44.2 ± 3.1), and higher in eyes open (-43.4 ± 2.6) versus eyes closed (-34.0 ± 2.6). The average phase difference between CO and EMG was significantly high in ULF (13.7 ± 4.8) versus VLF (-4.7 ± 4.8).

The CO signal lagged behind EMG significantly less in VLF (32.3 ± 3.0) versus LF (44.9 ± 3.0) and ULF (46.1 ± 3.0). The COPx lag was significantly high in LF (54.6 ± 2.6) than the VLF (29.5 ± 2.6) and ULF (29.8 ± 2.6) bands. Similarly COPx lead was significantly high in LF (-43.2 ± 6.3) than VLF (-26.0 ± 6.3) and ULF (-26.7 ± 6.3) bands. The average phase between COPy and EMG signals was low in the ULF (-13.6 ± 8.5) than LF (5.6 ± 8.5) and VLF (0.1 ± 8.5) bands, indicating a net phase lead of COPy over EMG. The COPy signal lagged less in VLF (31.4 ± 3.3) than LF (47.4± 3.3) and ULF (43.3 ± 3.3) bands, and led more in the VLF band (-30.1 ± 5.2) than LF (-46.8 ± 5.2) and ULF (-48.8 ± 5.2) bands.

The average values for the variables showed no effect of eye condition in all three-frequency bands (Table 3-3).

The SDA output (Table 3-4) showed significant association with the eyes conditions. The short-term scaling exponent Hys, was significantly low in the eyes open (0.09 ± 0.05) versus eyes closed (0.15 ± 0.05) condition; and Hrs was significantly low in eyes open (0.13 ± 0.05) than eyes closed (0.19 ± 0.05) condition.

The average value of significant coherence (> threshold) for the signal combinations is shown in table 3-5. The average coherence value was statistically different in all three frequency bands for the EMG – SBP (LF: 0.33 ± 0.01; VLF: 0.41 ± 0.01: ULF: 0.45 ± 0.01), EMG - DBP (LF: 0.33 ± 0.01; VLF: 0.41 ± 0.01; ULF: 0.44 ± 0.01) and EMG – CO (LF: 0.33 ± 0.01; VLF: 0.40 ± 0.01; ULF: 0.44 ± 0.01) combination. The average coherence for the EMG - COPx comparison was significantly high in ULF (0.56 ± 0.03) versus LF (0.47 ± 0.03) and VLF (0.49 ± 0.03) bands. Similarly, for EMG – COPy, the average coherence was significantly low in LF (0.39 ± 0.01) versus VLF (0.43 ± 0.01) and ULF (0.46 ± 0.01) bands, and low in eyes open (0.41 ± 0.01) versus eyes closed (0.45 ± 0.01) condition.

The average value of the variables in the time periods of significant coherence (> threshold) is shown in Table 3-6. No significant difference was detected in any of the variables across frequency bands. In COPy there was a significant difference with eyes condition in all frequency bands (ec: 0.09 ± 0.0; eo: 0.09 ± 0.01).

Table 3-7 shows the percentage of time of significant coherence over total 4 minute duration. For the EMG – SBP, EMG – DBP and EMG – CO combinations, the smallest median value occurs in the VLF and highest in LF frequency bands; for EMG – COPx and EMG – COPy combinations, the smallest median value occurs in ULF and highest in LF frequency bands.

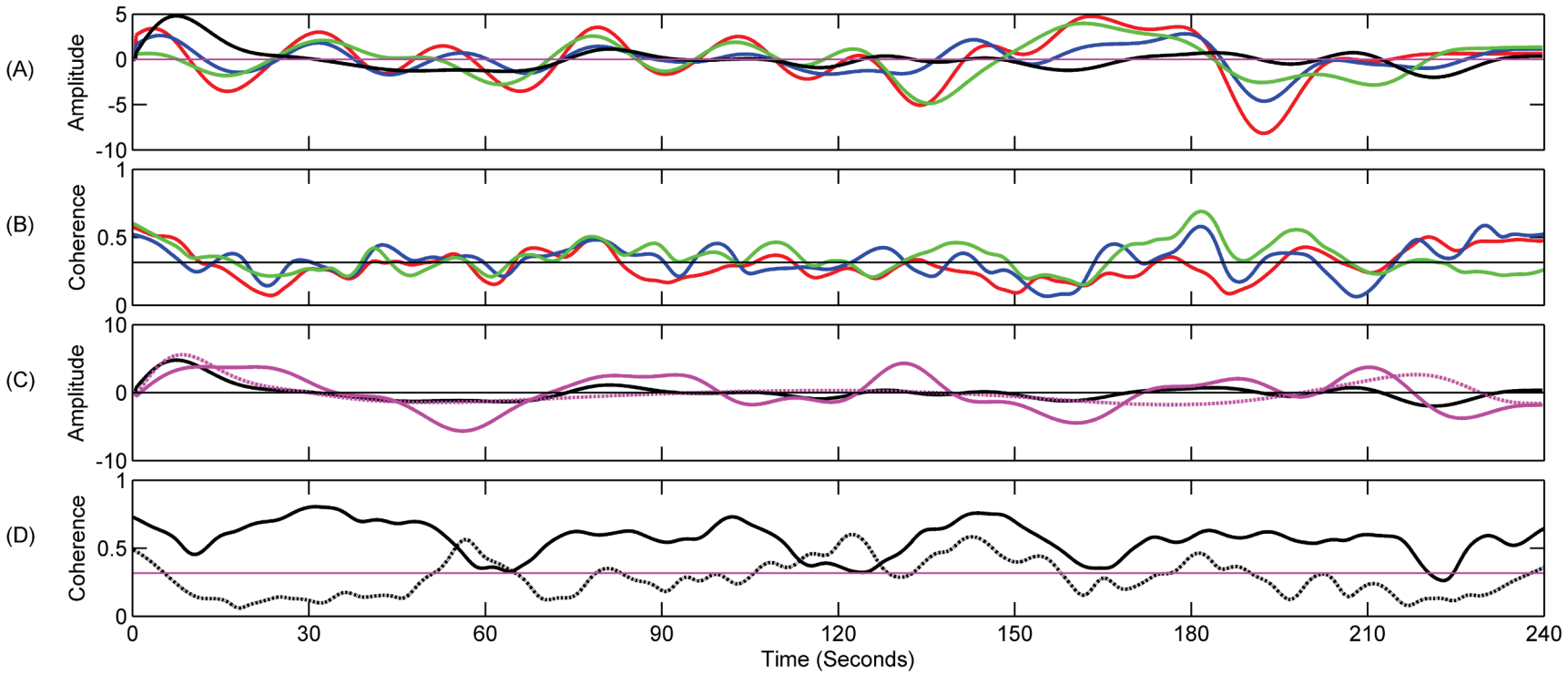

Figure 3-2 **Signals and the corresponding coherence time series for the VLF (0.01 – 0.05 Hz) frequency band. (A) EMG (black); SBP (red); DBP (blue); CO (green) signals (zero mean). (B) Coherence time series for the EMG-SBP (red), EMG-DBP (blue) and EMG-CO (green) combinations. Straight line represents the significance level. (C) EMG (black), COPx (magenta solid), COPy (magenta dash) signals (zero mean). (D)Coherence time series for the EMG-COPx (solid line) and EMG-COPy (dash line) combinations. Straight line represents the significance level. The zero mean EMG signal magnified by 2000X and CO signal magnified by 10X**

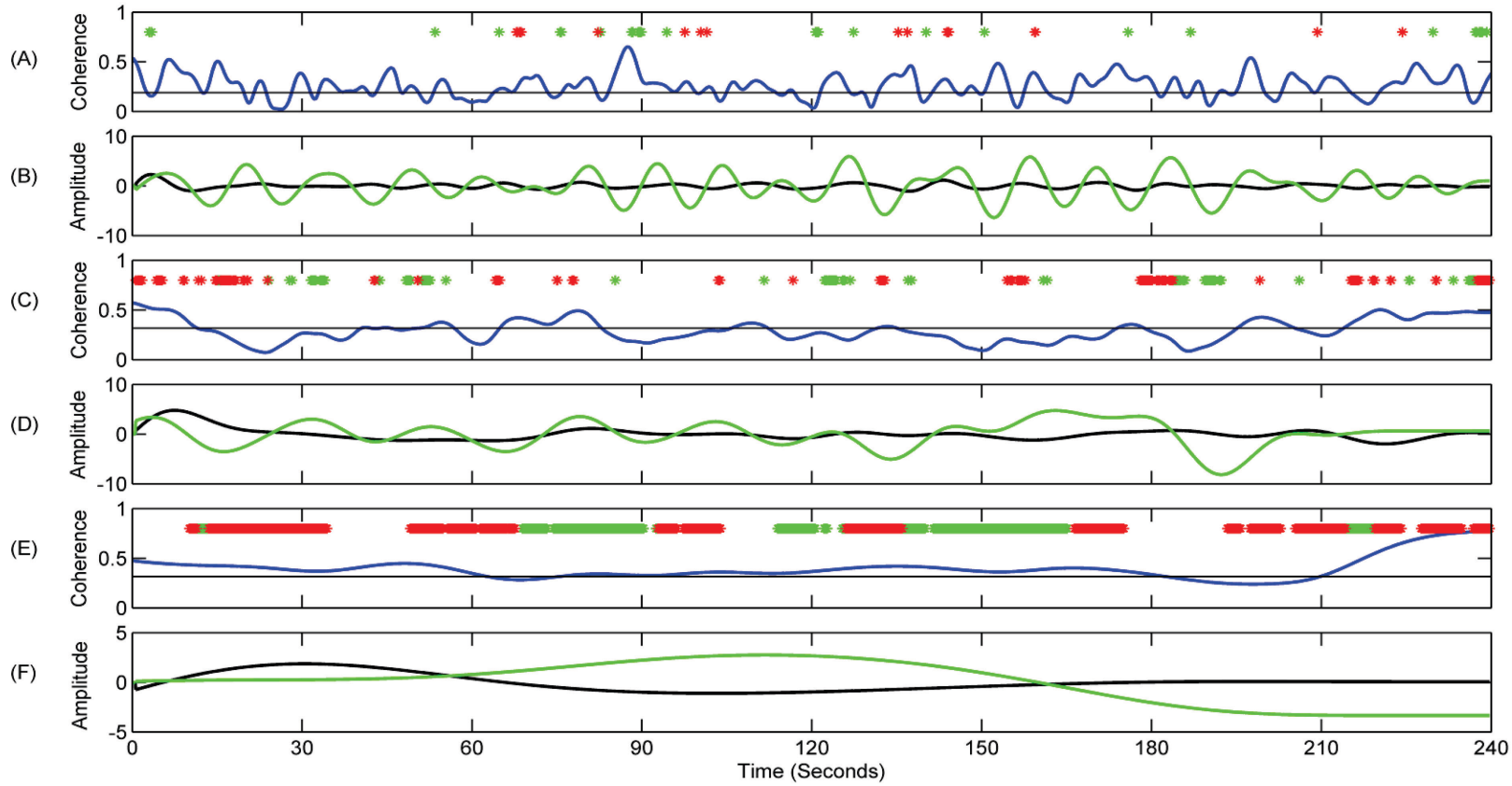

Figure 3-3 Coherence time series and the corresponding SBP (green) and EMG (black) signals for the entire duration of four minutes. The coherence time series and the corresponding signals are shown for the LF (0.05-0.1 Hz) (A & B), VLF (0.01 – 0.05 Hz) (C & D) and ULF (0.005 – 0.01 Hz) (E & F) plots. Phase lock depicted in the coherence plots as the asterisk (*) symbol. A red asterisk represents phase lead, and a green asterisk represents phase lag of the signal against the EMG signal. The straight line in the coherence plots represents the significance level for the corresponding frequency band. The zero mean EMG signal magnified by 2000X

Table 3-2 Least square mean value of the average phase angle between the variable and EMG in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). The averages include the time points for Phase lock, phase lock+ phase lag, phase lock + phase lead. Young participants, n = 5. Data reported as Mean (SEM).

	LF		VLF		ULF	
	EO	EC	EO	EC	EO	EC
SBP	5.69(6.49)	0.84(6.49)	1.61(6.49)	-9.46(6.49)	-6.12(6.49)	-12.17(6.49)
SBP Lag	45.08(4.90) €	35.43(4.90)	33.58(4.90) €	29.28(4.90)	44.09(4.90) €	32.60(4.90)
SBP Lead	-39.60(4.10)	-39.04(4.10)	-35.96(4.10)	-35.65(4.10)	-55.76(4.10) #	-45.58(4.10) #
DBP	-4.65(5.46)	3.58(5.46)	-1.75(5.46)	-4.90(5.46)	-11.35(5.46)	-11.61(5.46)
DBP Lag	44.13(4.63)	34.89(4.63)	27.72(4.63)	33.92(4.63)	45.20(4.63)	33.24(4.63)
DBP Lead	-48.98(4.38) € #	-30.35(4.38) #	-28.85(4.38) €	-36.07(4.38)	-52.62(4.38) €	-35.83(4.38)
CO	5.65(6.61)	0.76(6.61)	-8.78(6.61)	-0.56(6.61)	16.63(6.61) @	10.76(6.61) @
CO Lag	47.07(4.55)	42.65(4.55)	32.23(4.55) #	32.40(4.55) #	50.12(4.55)	42.10(4.55)
CO Lead	-37.82(5.03)	-31.09(5.03)	-37.01(5.03)	-34.71(5.03)	-37.50(5.03)	-36.15(5.03)

	LF		VLF		ULF	
	EO	EC	EO	EC	EO	EC
COPx	10.61(14.07)	12.97(14.07)	11.08(14.07)	-0.50(14.07)	9.15(14.07)	9.40(14.07)
COPx Lag	56.61(3.84) #	52.50(3.84) #	32.85(3.84)	26.06(3.84)	29.71(3.84)	29.87(3.84)
COPx Lead	-42.14(7.48) #	-44.33(7.48) #	-25.01(7.48)	-26.91(7.48)	-29.49(7.48)	-23.90(7.48)
COPy	4.85(9.65)	6.39(9.65)	0.27(9.65)	0.04(9.65)	-13.19(9.65) #	-14.09(9.65) #
COPy Lag	49.48(4.41)	45.29(4.41)	30.01(4.41) #	32.83(4.41) #	41.15(4.41)	45.45(4.41)
COPy lead	-46.58(6.55)	-46.92(6.55)	-29.47(6.55) #	-30.72(6.55) #	-46.75(6.55)	-50.86(6.55)

€: significantly different from the value in the eyes closed condition in all bands;
#: significantly different from the value in the other two frequency bands in all conditions.
@: significantly different from the value in the VLF band in all conditions.
Significance level at $p < 0.05$.

Table 3-3 Least square mean values for the average value for the variable in the phase lock (1), phase lock+ phase lag (2), phase lock + phase lead (3) portion of the whole time series in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). Phase lag and lead are with respect to the signal from the EMG signal. The transfer function values (A/B) were obtained for the phase lock portion only. Young participants, (n = 5). Data reported as Mean (SEM).

	LF		VLF		ULF	
	EO	EC	EO	EC	EO	EC
EMG/SBP	4.93 (0.6)E-05	5.06 (0.5)E-05	4.91 (0.6) E-05	5.09 (0.6)E-05	4.94 (0.6) E-05	5.06 (0.6) E-05
EMG/DBP	7.63(0.9)E-05	7.67 (0.9) E-05	7.70 (0.9) E-05	7.67(0.9) E-05	7.64 (0.9) E-05	7.67 (0.9) E-05
EMG/CO	1.43 (0.2) E-03	1.43 (0.2) E-03	1.42 (0.2) E-03	1.43 (0.2) E-03	1.39 (0.2) E-03	1.45 (0.2) E-03
EMG/COPx	-5.43 (7.51)E-01	4.87 (75.1)E-02	-5.48 (7.51)E-01	2.81 (75.1)E-02	-5.24 (7.51)E-01	-2.01 (75.1)E-02
EMG/COPy	2.48(10.5)	7.00 (10.5)	2.28 (10.5)	16.0 (10.5)	2.08 (10.5)	-22.7 (10.5)
EMG	5.76 (0.8) E-03	5.82 (0.8) E-03	5.73 (0.8) E-03	5.82(0.8) E-03	5.75 (0.8)E-03	5.82 (0.8) E-03
SBP	116(4)	113(4)	116(4)	113(4)	116(4)	113(4)
EMG	5.71(0.8) E-03	5.81 (0.8)E-03	5.74(0.8) E-03	5.67(0.8) E-03	5.73 (0.8) E-03	5.73 (0.8) E-04
SBP Lag	116(4)	113(4)	116(4)	112(4)	116(4)	114(4)
EMG	5.83 (0.8) E-03	5.82(0.8) E-03	5.68 (0.8) E-03	5.93 (0.8) E-03	5.77 (0.8) E-03	5.88 (0.8) E-03
SBP Lead	116(4)	113(4)	116(4)	113(4)	117(4)	113(4)

	LF		VLF		ULF	
	EO	EC	EO	EC	EO	EC
EMG	5.72 (0.8) E-03	5.82 (0.8) E-03	5.77(0.8) E-03	5.85 (0.8)E-03	5.70(0.8) E-03	5.84 (0.8)E-03
DBP	75(3)	75(3)	75(3)	75(3)	75(3)	75(3)
EMG	5.69 (0.8) E-03	5.83 (0.8) E-03	5.74 (0.8) E-03	5.78 (0.8) E-03	5.76 (0.8) E-03	5.73 (0.8) E-03
DBP Lag	75(3)	75(3)	75(3)	75(3)	74(3)	75(3)
EMG	5.75 (0.8) E-03	5.79 (0.8) E-03	5.80 (0.8) E-03	5.91 (0.8) E-03	5.64 (0.8) E-04	5.91 (0.8) E-03
DBP Lead	75(3)	75(3)	75(3)	75(3)	75(3)	75(3)
EMG	5.73 (0.8) E-03	5.83 (0.8) E-03	5.77 (0.8) E-03	5.82 (0.8) E-03	5.64 (0.8) E-03	5.84 (0.8) E-03
CO	4.07(0.30)	4.19(0.30)	4.10(0.30)	4.18(0.30)	4.14(0.30)	4.13(0.30)
EMG	5.73 (0.8) E-03	5.86 (0.8) E-03	5.71 (0.8) E-03	5.79 (0.8) E-03	5.65 (0.8) E-03	5.79 (0.8) E-03
CO Lag	4.06(0.29)	4.20(0.29)	4.08(0.29)	4.17(0.29)	4.16(0.29)	4.10(0.29)
EMG	5.74 (0.8) E-03	5.83 (0.8) E-03	5.80 (0.8) E-03	5.85 (0.8) E-03	5.60 (0.8) E-03	5.92 (0.8) E-03
CO Lead	4.08(0.30)	4.18(0.30)	4.12(0.30)	4.20(0.30)	4.12(0.30)	4.18(0.30)
EMG	5.78 (0.8) E-03	5.82 (0.8) E-03	5.78 (0.8) E-03	5.84 (0.8) E-03	5.71 (0.8) E-03	5.87 (0.8) E-03

	LF		VLF		ULF	
	EO	EC	EO	EC	EO	EC
COPx	-1.66 (1.17)E-02	-1.53 (1.17)E-02	-1.65 (1.17)E-02	-1.49 (1.17)E-02	-1.69 (1.17)E-02	-1.48 (1.17)E-02
EMG	5.77 (0.8) E-03	5.81 (0.8) E-03	5.74 (0.8) E-03	5.80 (0.8) E-03	5.56 (0.8) E-03	5.82 (0.8) E-03
COPx Lag	-1.67 (1.18)E-02	-1.53 (1.18)E-02	-1.65 (1.18)E-02	-1.49 (1.18)E-02	-1.74 (1.18)E-02	-1.45 (1.18)E-02
EMG	5.83 (0.8) E-03	5.74 (0.8) E-03	5.82 (0.8) E-03	5.85 (0.8) E-03	6.04 (0.8) E-03	6.07 (0.8) E-03
COPx Lead	-1.61 (1.15)E-02	-1.52 (1.15)E-02	-1.50 (1.15)E-02	-1.52 (1.15)E-02	-1.57 (1.15)E-02	-1.54 (1.15)E-02
EMG	5.75 (0.8) E-03	5.79 (0.8) E-03	5.78 (0.8) E-03	5.85 (0.8) E-03	5.69 (0.8) E-03	5.87 (0.8) E-03
COPy	8.95 (8.94)E-02	9.04 (8.94)E-02	8.96 (8.94)E-02	9.04 (8.94)E-02	8.99 (8.94)E-02	9.01 (8.94)E-02
EMG	5.76 (0.8) E-03	5.79 (0.8) E-03	5.77 (0.8) E-03	5.70 (0.8) E-03	5.56 (0.8) E-03	5.74 (0.8) E-03
COPy Lag	8.95 (8.94)E-02	9.04 (8.94)E-02	9.02 (8.94)E-02	8.99 (8.94)E-02	8.98 (8.94)E-02	9.08 (8.94)E-02
EMG	5.75 (0.8) E-03	5.82 (0.8) E-03	5.78 (0.8) E-03	5.99 (0.8) E-03	5.78 (0.8) E-03	5.98 (0.8) E-03
COPy Lead	8.95 (8.94)E-02	9.04 (8.94)E-02	8.93 (8.94)E-02	9.04 (8.94)E-02	8.98 (8.94)E-02	9.00 (8.94)E-02

Table 3-4 Least square mean value of the output from the SDA analysis applied to the COP data from the young age group (n = 12). Data reported as Mean (SEM).

	M		F	
	EO	EC	EO	EC
delta txc	0.807(0.032)	0.811(0.033)	0.713(0.032)	0.720(0.029)
X2	1.28 (0.6) E-04	1.38 (0.6) E-04	0.9(57.1) E-06	0.9 (57.1) E-06
Dxs	8.04 (3.87) E-05	9.42 (3.88) E-05	1.41 (36.8) E-06	1.70 (36.7) E-06
Dxl	7.14 (3.69) E-05	7.41 (3.69) E-05	4.02E-21(3.51E-05)	2.75E-21(3.50E-05)
Hxs	0.210(0.059)	0.229(0.060)	-0.023(0.058)	0.082(0.055)
Hxl	0.242(0.046)	0.222(0.046)	0.116(0.044)	0.106(0.043)
delta tyc	0.811(0.032)	0.810(0.033)	0.712(0.032)	0.716(0.029)
Y2	1.30 (0.6) E-04	1.38 (0.6) E-04	0.86 (57.6) E-06	1.00 (57.6) E-06
Dys	8.02 (3.86) E-05	9.41 (3.86) E-05	1.51 (36.7) E-06	1.80 (36.6) E-06
Dyl	7.14 (3.69) E-05	7.41 (3.69) E-05	4.02E-21(3.51E-05)	2.75E-21(3.50E-05)

	M		F	
	EO	EC	EO	EC
Hys	0.197(0.059)	0.227(0.061)	-0.025(0.059)	0.081(0.055)
Hyl	0.241(0.046) #	0.222(0.046)	0.115(0.044) #	0.107(0.043)
Delta trc	0.809(0.031)	0.809(0.032)	0.712(0.032)	0.716(0.029)
R2	2.57 (1.21) E-04	2.73 (1.21) E-04	0.15 (11.4) E-05	0.20 (11.4) E-05
Drs	1.61 (0.8) E-04	1.88 (0.8) E-04	2.48 (73.4) E-06	3.20 (73.2) E-06
Drl	1.42 (0.7) E-04	1.48 (0.7) E-04	1.96E-20(7.01E-05)	2.11E-20(7.00E-05)
Hrs	0.238(0.057) #	0.264(0.059)	0.026(0.057) #	0.124(0.053)
Hrl	0.242(0.046)	0.221(0.046)	0.116(0.044)	0.106(0.043)
#: significantly different from the value in the eyes closed condition in both genders Significance level at $p<0.05$.				

Table 3-5 Least square mean of the average coherence value in the coherence> threshold regions for young participants in ec/eo conditions in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). (n = 5). The coherence estimate is calculated between the variable and EMG with time synchronization. Data reported as Mean (SEM).

	LF		VLF		ULF	
	EO	EC	EO	EC	EO	EC
SBP	0.350(0.015) #	0.314(0.015) #	0.411(0.015) #	0.411(0.015) #	0.453(0.015) #	0.447(0.015) #
DBP	0.342(0.013) #	0.324(0.013) #	0.407(0.013) #	0.411(0.013) #	0.454(0.013) #	0.446(0.013) #
CO	0.343(0.015) #	0.317(0.015) #	0.398(0.015) #	0.408(0.015) #	0.447(0.015) #	0.436(0.015) #
COPx	0.453(0.035)	0.485(0.035)	0.472(0.035)	0.502(0.035)	0.577(0.035) #	0.549(0.035) #
COPy	0.380(0.017) # @	0.408(0.017) #	0.409(0.017) @	0.454(0.017)	0.431(0.017) @	0.481(0.017)

#: significantly different from the values in other two frequency bands.;
@: significantly different from the value in the eyes closed condition ;
Significance level at $p<0.05$

Table 3-6 Least square mean of the average value of the variables in the regions of significant coherence for young participants in ec/eo conditions in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz) (n = 5). Data reported as Mean (SEM).

	LF		VLF		ULF	
	EO	EC	EO	EC	EO	EC
EMG	5.76 (0.8) E-03	5.82(0.8) E-03	5.68 (0.8) E-03	5.79 (0.8) E-03	5.71 (0.8) E-03	5.87 (0.8) E-03
SBP	116(4)	113(4)	116(4)	113(4)	116(4)	113(4)
EMG 2	5.77 (0.8) E-03	5.80 (0.8) E-03	5.72 (0.8) E-03	5.76 (0.8) E-03	5.72 (0.8) E-03	5.89 (0.8) E-03
DBP	75(3)	75(3)	75(3)	75(3)	74(3)	75(3)
EMG 3	5.76 (0.8) E-03	5.81 (0.8) E-03	5.70 (0.8) E-03	5.91 (0.8) E-03	5.73 (0.8) E-03	5.81 (0.8) E-03
CO	4.08(0.30)	4.17(0.30)	4.10(0.30)	4.17(0.30)	4.20(0.30)	4.17(0.30)
EMG 4	5.76 (0.8) E-03	5.80 (0.8) E-03	5.77 (0.8) E-03	5.83 (0.8) E-03	5.76 (0.8) E-03	5.83 (0.8) E-03
COPx	-1.67(1.17)E-02	-1.51(1.17)E-02	-1.66(1.17)E-02	-1.50(1.17)E-02	-1.65(1.17)E-02	-1.49(1.17)E-02
EMG 5	5.77 (0.8) E-03	5.79 (0.8) E-03	5.80 (0.8) E-03	5.81 (0.8) E-03	5.72 (0.8) E-03	5.78 (0.8) E-03
COPy	8.94 (8.93)E-02 #	9.04 (8.93)E-02	8.92 (8.93)E-02 #	9.05 (8.93)E-02	8.96 (8.93)E-02 #	8.99 (8.93) E-02
#: significantly different from the value in the eyes closed condition; Significance level at $p<0.05$						

Table 3-7 Percentage of time of significant coherence over total time of 4 minutes. LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz)Young participants (n = 13), eyes closed condition.

		MEDIAN		MIN		MAX
SBP	LF	61.8		48.9		74.8
	VLF	39.9		16.5		65.5
	ULF	45.8		13.3		83.2
DBP	LF	62.4		44.3		73.0
	VLF	42.6		17.6		69.9
	ULF	51.3		14.2		73.8
CO	LF	64.1		49.5		78.1
	VLF	49.2		24.1		59.8
	ULF	53.2		22.1		81.1
COPx	LF	88.3		81.8		99.7
	VLF	73.9		34.8		98.0
	ULF	73.6		37.0		99.7
COPy	LF	83.0		70.3		97.2
	VLF	68.9		36.3		93.5
	ULF	60.0		29.4		95.3

3.5 Conclusion

The postural and cardiovascular systems were investigated for a possible interdependence. The representative variables, systolic and diastolic blood pressure, cardiac out, posture muscle electromyography, and center of pressure, were analyzed to find the existence of a relationship.

The average values of coherence where it was significant (>significance threshold) for the cardiovascular variables with EMG combination showed a statistically significant ($p<0.05$) association with frequency bands, indicating a frequency dependent interaction between the cardiovascular and postural systems. The average coherence between the COP trajectory in the antero-posterior direction showed a dependence on the eyes condition. There was an increased coherence, and, hence, an increased coupling between EMG and COP with no visual input which implied a reliance on neuromuscular control rather than central control of posture.

The statistical comparison of the SDA analysis outcome showed no change in the diffusion coefficient values, indicating that the degree of randomness in the COP trajectory does not change with the removal of visual input and has no association with gender. The short-term scaling exponent showed an increase with the eyes closed condition in the mean displacement and the antero-posterior direction, suggesting that the tendency to sway back to the equilibrium, or mean position, was reduced with the removal of visual input. Hence the tendency to rely on the closed loop control reduced with the removal of visual input.

The average values of the phase angle difference between all variables and EMG showed no consistent behaviour across all frequency bands. The systolic and diastolic blood pressure lag from EMG was reduced with eyes closed, implying a faster response of blood pressure to an EMG event. Phase lead was significantly high in the ULF band, indicating that blood pressure events occur ahead of the muscle contractions in a slow time pattern.

The average value for the variables showed no association with the eyes condition across all frequency bands, suggesting no effect in the young age group of deprivation from visual input.

In summary, the cardiovascular and the postural systems show the existence of a bi-directional relationship. The posture muscles work to regain posture and pump blood back to the heart. Elimination of visual input to the central nervous system elevated the postural control system activity with short-term strategy, and increased the activation of the posture muscles. The action of the muscles as a skeletal muscle pump was dominant in the ultra low frequency range for the young participants. This could indicate its possible function in a long-term stabilization approach.

CHAPTER 4: EFFECTS OF AGEING ON THE INTERRELATIONSHIP BETWEEN THE CARDIO – POSTURAL CONTROL

4.1 Abstract

The scientific community has long identified physiological changes with age. Due to these changes, there are notable vulnerabilities, which include falls, fractures and proneness to cardiovascular diseases. Studies have indicated (Chapter 3) that there is a bidirectional relationship between the postural and cardiovascular controls. This relationship was hypothesized to be an effected by ageing. Wavelet methods described in chapter 3 were applied to the comparison of cardiovascular and postural control with gender and age. Stabilogram diffusion analysis was applied to find the effect of ageing on postural control. This study found the cardiovascular and postural control relationship to have an association with age and gender. Elevated levels of systolic blood pressure and calf muscle EMG activity were observed in the elderly participant group.

4.2 Introduction

One in three persons over 65 years old and one in two aged 85 or over will fall each year; about half of these falls involve bone fractures (Wilkins, 1999). Falls are a leading cause of injury related morbidity and mortality in the elderly, and are the sixth leading cause of death. Often, these falls result from a loss of balance due to decreased postural control. Age related reductions in

cardiovascular autonomic control and/or autonomic neuropathy can lead to hypotension during postural changes, and may produce light-headedness and fainting (syncope). Further study is required to increase our knowledge and better characterize how the cardiovascular and the postural control systems interact in order to assess fall proneness.

It has been reported in previous research that the percentage of skeletal muscle mass, relative to body mass, is higher in males (38.4%) than females (30.6%) (Janssen et al 2000). This may have an effect on the amount of venous return during upright posture in females. The maintenance of upright posture not only requires coordinated vestibular-neuromuscular control of postural muscles, but also cardiovascular reflexes to maintain blood pressure.

Previous research indicates a reduction in skeletal muscle mass, irrespective of gender, with ageing (Janssen et al 2000). Lower extremity muscle strength has been shown to reduce as much as 40% between 30 and 80 years of age (Aniansson et al 1986).

The ability to minimize spontaneous sway has also been shown to be reduced in the elderly (age 50 to 80 years) (Sheldon 1963), with the greatest amount of spontaneous sway observed in the 80 year olds (Toupet et al 1992). Bruce (1980) observed a decline in the tactile sensitivity over age. Fine touch, vibration and pressure sensitivities have also been observed to reduce over age, as there is a documented reduction of fibers innervating peripheral receptors. It has been exhibited in many experiments that there is an increase in the overall sway when vision is removed (Patla et al 1990). The vestibular system, which

acts as an absolute reference for posture control (Black and Nashner 1985), has a reduced function with a loss of vestibular hair and nerve cells with age (Rosenhall and Rubin 1975).

An age related decline in vascular elasticity, loss of venous valve function, and reduced baroreceptor sensitivity causes individuals to be prone to orthostatic intolerance resulting in falls due to postural related syncope. The vasovagal syncope is associated with reduced heart rate in conjunction with vasodilation while vasodepressor syncope is related to a fall in vascular resistance. Without an active muscle pump or a reduced activity of the pump, central venous pressure falls, followed by a reduction in cardiac output and finally blood pressure (Rowell 1993). If this decline remains uncompensated by other alternative measures, fainting will occur.

With a reduction in the amount of blood returned to central circulation and venous pooling occurring in lower extremities due to gravitational effects, it was hypothesized that there would be an increase in postural muscle EMG signal amplitude in the elderly age group compared to the young age group participants. This increase in EMG amplitude would be attributed to both posture correction (increased postural sway) and skeletal muscle pump (compensation for blood pooling). The healthy segment of the elderly population was studied in order to identify the effect of physiological changes with age on the relationship. The baseline characteristics for the elderly identified in the present analysis, can be used further to study the fallers and other individuals with diseases.

4.3 Methods

4.3.1 Data Analysis

The WTC estimate was determined for the signal combinations under consideration using the method defined in section (2.3.1). The same value of ω_0 =6 was used for all the WTC estimate calculations. Phase difference and phase lock was found for all the signal combinations as in chapter 3 (section 3.3.1.1).

To test whether the outcome variable values were associated with gender (2 levels) and age (2 levels) a repeated measure ANOVA analysis was conducted. The significance level for the tests was kept at α = 0.05, and all analyses were conducted with the JMP 7 statistical package (SAS Inc, USA).

4.3.2 Data collection

Additional verbal/ visual screening was performed prior to testing to identify any conditions that may have been left unnoticed/undocumented during the medical pre-screening. Experimental data were collected with the same protocol as followed for the young individuals and defined in section (3.3.2)

4.3.3 Participant pool

Data were collected from 15 participants (8 females and 7 males). Age was limited between 65 to 75 years of age. Participants were recruited from local senior centers, walking groups and the local SFU community. All volunteers completed a medical screening form and their family physician was consulted to ensure a medical history devoid of: a) any cardiovascular complication; b) any

cardiovascular implants; c) any neurological disorders; d) undergone any orthopaedic surgery in the lower limbs in the past 2 years; e) unexplained falls in the past 12 months; and, f) prescription medications for heart or brain function. Written approval was obtained from all participants.

4.4 Results

4.4.1 Analysis for elderly participant group

Data for one male participant was not included in the analysis as the blood pressure and ECG waveforms had a regular appearing ectopic beat. The disruption of the normal rhythm was followed by erroneous measurement of the cardiovascular variables. The total number of participants was therefore reduced to 14. Table 4-1 below enumerates the anthropometric data for the participant group.

Table 4-1 Anthropometric data for elderly participant group.

S. No.	Participant No.	Gender	Age (years)	Height (cm.)	Weight (kg.)
1	1	F	68	153	47.5
2	2	F	67	169	71
3	3	F	68	152.2	76
4	4	M	80	160	68
5	5	F	69	162.5	57
6	6	F	70	167.5	61
7	7	M	66	172	101
8	8	M	65	187	64
9	9	F	67	146	44
10	10	F	66	157.5	50
11	12	M	69	169	83
12	13	M	66	178	74
13	14	F	72	152	44
14	15	M	69	185	83

Figure 4-1a & b demonstrate segments when the signals for SBP, DBP and CO behave in a similar manner, and as a result, have a significant coherence at the same time point. Additionally, around the time of 150 seconds, the CO signal has significant coherence and correspondingly shows an increase in amplitude. In Figure 4-1 c & d, the COPx and COPy signals show a rapid change in behaviour, resulting in short segments of significant coherence.

The representative participant (participant No. 9) shows a peculiar attribute near zero significant coherence in the ULF band, and a dominant coherence in the LF band (Figure 4-2). The phase locking is prominent in the ULF band, with changing behaviour from lead to lag. As there are many peaks in the coherence plot in the LF region, the phase locking periods consequently shift in behaviour from phase lead to lag to lead, which indicates a catch-up process between the two signals.

The average value of the phase angle (Table 4-2) of SBP was highest in ULF (13.1 ± 2.8) than LF (-1.7 ± 2.8) and VLF (-5.96 ± 2.8); SBP lead was lowest in VLF (31.7 ± 2.2) than LF (44.85 ± 2.2) and ULF (48.17 ± 2.2); and SBP lag was lowest in VLF (-33.4 ± 2.2) than LF (-44.3 ± 2.2) and ULF (-44.7 ± 2.2). The average phase angle of DBP was highest in ULF (9.5 ± 2.8) than LF (1.4 ± 2.8) and VLF (-2.43 ± 2.8); DBP lag was significantly different in all three bands LF (42.7 ± 1.7); VLF (31.2 ± 1.7), ULF (48.9 ± 1.7), and DBP lead was lowest in VLF (-33.2 ± 2.3) than LF (-42.3 ± 2.3) and ULF (-46.7 ± 2.3).

The average value for the CO lag was lowest in VLF (33.1 ± 2.0) than LF (39.5 ± 2.0) and ULF (42.5 ± 2.0); CO lead was lowest in VLF (-34.7 ± 2.2) which was significantly different from the value in ULF (-44.9 ± 2.2) band.

The average value of phase angle for COPx was significantly different in LF (-40.8 ± 3.9); VLF (-14.8 ± 3.9) and ULF (7.5 ± 3.9). Males had considerably low values of average angle in LF (-49.9 ± 5.9) and VLF (-23.40 ±5.9) bands and high value in ULF (12.51 ± 5.9) band. Mean value of phase angle for COPx lag was lowest in VLF (28.8 ± 2.4) and significantly from other bands (LF: 37.4 ± 2.4; ULF: 37.0 ± 2.4). The COPx lead mean value of phase angle was lowest in LF (-63.1 ± 2.0) and significantly different from other bands (VLF: -38.9 ± 2.0; ULF: -32.6 ± 2.0). The mean value of the phase angle for males was significantly different from females in LF (M: -68.6 ± 3.1, F: -57.6 ± 2.7, $p<0.05$) and VLF (M: -43.1 ± 3.1, F: -34.6 ± 2.72, $p<0.05$) bands.

The mean value of the phase angle for the COPy was significantly different (VLF: 8.0 ± 3.6; ULF: -0.24 ± 3.6) and highest in LF (33.6 ± 3.6) band; significantly high in eyes closed (17.5 ± 3.0) than eyes open (10.1 ± 3.0). Phase angle values for COPy lag was significantly different and high in males (49.6 ± 1.8, F: 43.6 ± 1.5) and significantly different across three bands (LF: 56.6 ± 1.8, VLF: 32.4 ± 1.8, ULF: 50.8 ± 1.8).

Phase angle values for COPy lead were significantly different in all three frequency bands (LF: -36.4 ± 2.1, VLF: -26.4 ± 2.1, ULF: -43.5 ± 2.1, $p<0.05$).

The average value of the transfer characteristics (Table 4-3) for all variables with EMG were not affected by gender, frequency or eyes

open/closed. The mean of SBP during the phase lock, lead and lag portions was significantly higher in the eyes closed (142 ± 7) versus the eyes open condition (138 ± 7). The cardiac output in phase lock, lag and lead case was significantly higher in males (8.49 ± 0.6) versus females (6.12 ± 0.5) and eyes closed (7.46 ± 0.4) versus eyes open (7.16 ± 0.4) condition. Mean value of the COPx coordinate was significantly lower in eyes closed ((3.80 ± 0.9) E-03) to eyes open ((3.97 ± 0.9) E-03) condition

The average value of significant coherence (> threshold) revealed a strong association with the frequency bands for all combinations (Table 4-4). The average value of coherence for EMG – SBP (LF: 0.33± 0.01; VLF: 0.40 ± 0.01; ULF: o.44 ± 0.01), EMG - DBP (LF: 0.34 ± 0.01; VLF: 0.40 ± 0.01; ULF: 0.46 ± 0.01), and EMG – CO (LF: 0.32 ± 0.01; VLF: 0.41 ± 0.01; ULF: 0.44 ± 0.01) were significantly different in all three frequency bands. Average coherence for EMG-COPx combination was significantly higher in males (0.50 ± 0.01) versus females (0.45 ± 0.01) and significantly higher in ULF (0.51 ± 0.01) than in LF (0.46 ± 0.01) and VLF (0.45 ± 0.01) bands. Similarly, EMG-COPy combination had significantly higher coherence in ULF (0.47 ± 0.01) than LF (0.44 ± 0.01) and VLF (0.44 ± 0.01) bands.

The average value of variables in the time periods of significant coherence (> threshold) showed no effect on EMG values in any condition. Both SBP (ec: 142 ± 7; eo: 138 ± 7) and CO (ec: 7.45 ± 0.4; eo: 7.15 ± 0.4) were significantly higher in eyes closed condition with no effect on DBP. CO was significantly higher in males (8.48 ± 0.6) than females (6.13 ± 0.5). COPx was significantly

high in eyes open ((3.9 ± 0.9) E-03) versus eyes closed ((3.8 ± 0.9) E-03) condition.

The output variable values from the SDA analysis (Table 4-6) in the elderly participant group showed no consistent effect of any condition. However, the long term scaling exponent in all three directions was significantly higher in the eyes open (0.14 ± 0.03) versus eyes closed (0.11 ± 0.03) condition.

Table 4-7 shows the percentage of time of significant coherence over total 4 minute duration. For the EMG – SBP, EMG – DBP and EMG – CO combinations, the smallest median value occurs in the VLF and highest in LF frequency bands; for EMG – COPx and EMG – COPy combinations, the smallest median value occurs in ULF and highest in LF frequency bands.

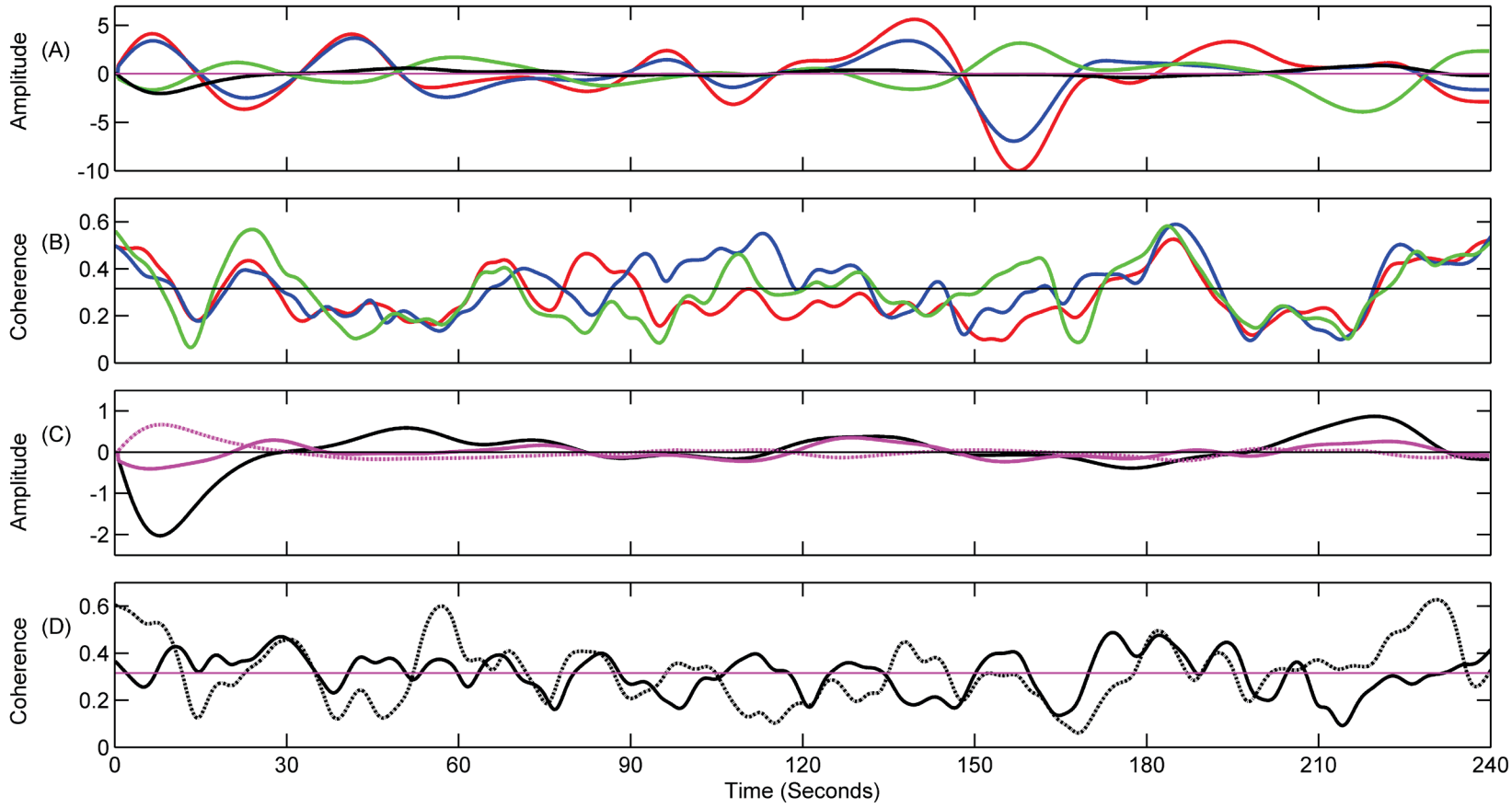

Figure 4-1 Signals and the corresponding coherence time series for the VLF (0.01 – 0.05 Hz)frequency band for the elderly participant. (A) EMG (black); SBP (red); DBP (blue); CO (green) signals (zero mean). (B) Coherence time series for the EMG-SBP (red), EMG-DBP (blue) and EMG-CO (green) combinations. Straight line represents the significance level. (C) EMG (black), COPx (magenta solid), COPy (magenta dash) signals (zero mean). (D) Coherence time series for the EMG-COPx (solid line) and EMG-COPy (dash line) combinations. Straight line represents the significance level. Zero mean EMG signal magnified by 2000X and CO signal magnified by 10X

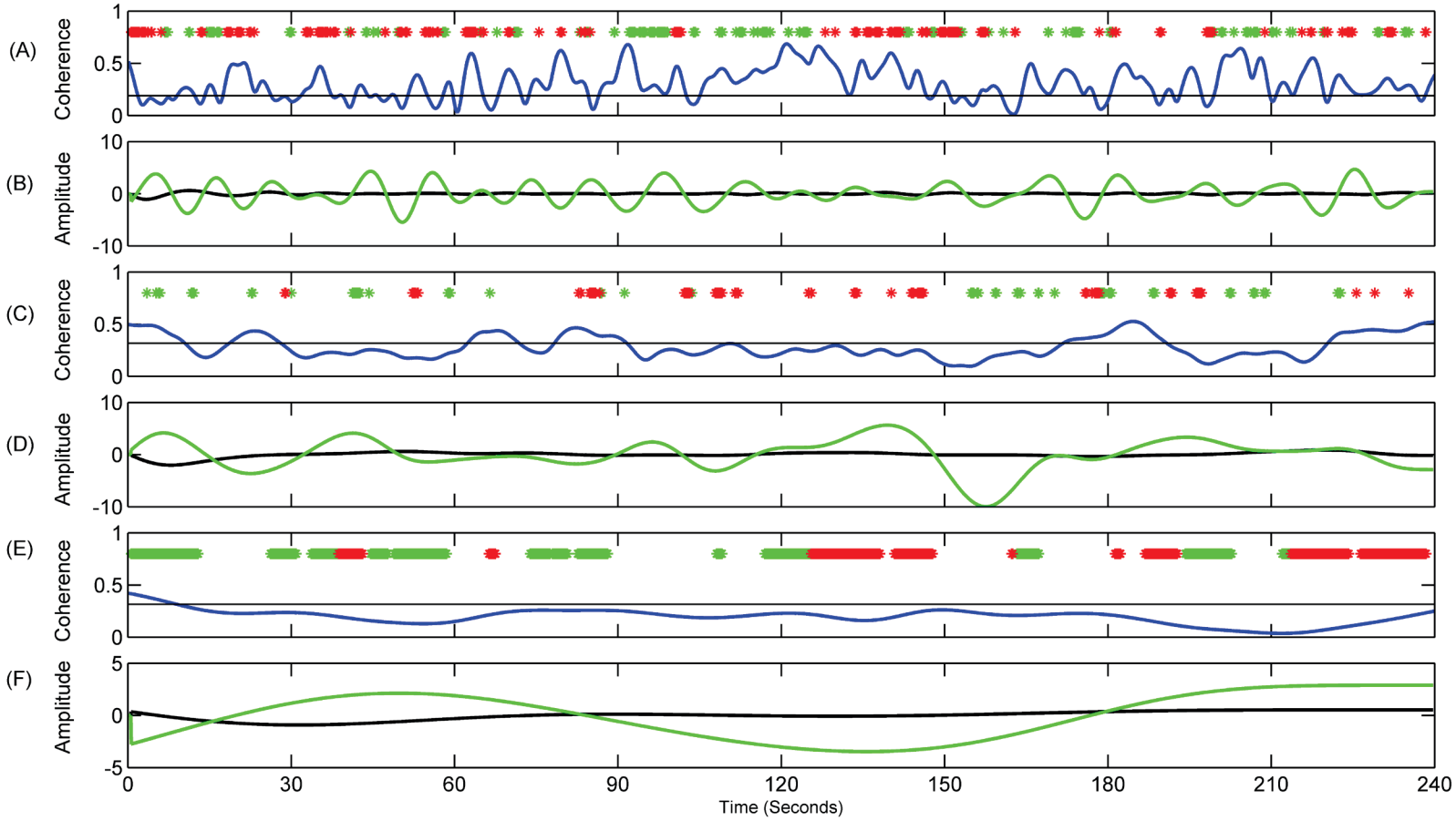

Figure 4-2Coherence time series and the corresponding SBP (green) and EMG (black) signals for the entire duration of four minutes for a representative elderly participant. The coherence time series and the corresponding signals are shown for the LF (0.05 – 0.1 Hz)(A & B), VLF (0.01 – 0.05 Hz) (C & D) and ULF (0.005 – 0.01 Hz) (E & F) plots. The phase locking is depicted in the coherence plots as the asterisk (*) symbol. A red asterisk represents phase lead and a green asterisk represents phase lag of the signal against the EMG signal. The straight line in the coherence plots represents the significance level for the corresponding frequency band. Zero mean EMG signal magnified by 2000X

Table 4-2 Least square mean values of average phase angle between EMG and the corresponding signal, under both eyes closed/open condition in elderly male and female participants for the three frequency bands. LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). Value calculated in the phase lock, phase lock + phase lag and phase lock + phase lead condition. (n = 14). Data reported as Mean (SEM)

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
SBP	-4.7 (5.2)	-2.6 (5.2)	0.3 (6.0)	0.1 (6.0)	-2.6 (5.2)	-4.9 (5.2)	-10.9 (6.0)	-5.5 (6.0)	18.0 (5.2) **	10.0 (5.2) **	17.7 (6.0) **	6.8 (6.0) **
SBP Lag	46.3 (4.1)	40.9 (4.1)	50.6 (4.7)	41.6 (4.7)	33.1 (4.1) **	32.0 (4.1) **	27.1 (4.7) **	34.6 (4.7) **	49.1 (4.1)	46.4 (4.1)	52.9 (4.7)	44.3 (4.7)
SBP Lead	-50.9 (4.1)	-39.7 (4.1)	-44.6 (4.8)	-41.8 (4.8)	-32.5 (4.1) **	-32.0 (4.1) **	-35.5 (4.8) **	-33.7 (4.8) **	-47.4 (4.1)	-41.2 (4.1)	-45.8 (4.8)	-44.3 (4.8)
DBP	0.9 (5.0)	-5.4 (5.0)	6.1 (5.8)	4.1 (5.8)	-2.9 (5.0)	-5.1 (5.0)	-2.0 (5.8)	0.3 (5.8)	16.0 (5.0) **	11.4 (5.0) **	8.4 (5.8) **	2.4 (5.8) **
DBP Lag	43.4 (3.0) **	38.6 (3.0) **	42.0 (3.5) **	47.1 (3.5) **	31.6 (3.0) **	28.7 (3.0) **	30.7 (3.5) **	34.0 (3.5) **	49.1 (3.0) **	47.4 (3.0) **	54.0 (3.5) **	45.2 (3.5) **
DBP Lead	-43.1 (4.1)	-46.9 (4.1)	-35.6 (4.7)	-43.9 (4.7)	-32.7 (4.1) **	-34.4 (4.1) **	-34.3 (4.7) **	-31.5 (4.7) **	-35.5 (4.1) #	-48.2 (4.1) #	-51.6 (4.7) #	-51.3 (4.7) #
CO	2.3 (5.4)	-6.2 (5.4)	2.7 (6.2)	-2.3 (6.2)	-5.5 (5.4)	0.4 (5.4)	-4.9 (6.2)	-2.3 (6.2)	-2.2 (5.4)	5.6 (5.4)	-6.8 (6.2)	7.8 (6.2)
CO Lag	39.4 (3.8)	38.2 (3.8)	40.7 (4.4)	39.7 (4.4)	33.7 (3.8) **	36.4 (3.8) **	30.2 (4.4) **	32.1 (4.4) **	38.2 (3.8)	51.2 (3.8)	36.4 (4.4)	44.0 (4.4)

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
CO Lead	-36.1 (4.1) **	-38.7 (4.1) **	-36.9 (4.7) **	-43.1 (4.7) **	-34.9 (4.1)	-33.0 (4.1)	-36.7 (4.7)	-34.0 (4.7)	-42.3 (4.1)	-46.3 (4.1)	-40.8 (4.7)	-50.0 (4.7)
COPx	-30.7 (6.4) **	-32.7 (6.4) **	-48.6 (7.4) ** #	-51.2 (7.4) ** #	-7.1 (6.4) **	-5.1 (6.4) **	-27.4 (7.4) ** #	-19.4 (7.4) ** #	0.2 (6.4) **	4.7 (6.4) **	5.2 (7.4) **	19.9 (7.4) **
COPx Lag	41.4 (4.4)	35.5 (4.4)	40.6 (5.0)	32.0 (5.0)	30.3 (4.4) **	33.7 (4.4) **	25.5 (5.0) **	25.8 (5.0) **	34.4 (4.4)	34.6 (4.4)	37.1 (5.0)	42.0 (5.0)
COPx Lead	-57.4 (3.5) **	-57.8 (3.5) **	-68.7 (4.0) ** #	-68.6 (4.0) ** #	-36.6 (3.5)	-32.7 (3.5)	-43.2 (4.0) #	-43.1 (4.0) #	-31.1 (3.5)	-38.2 (3.5)	-36.3 (4.0)	-25.0 (4.0)
COPy	26.7 (6.3) ** @	28.6 (6.3) **	36.3 (7.2) ** @	42.8 (7.2) **	5.6 (6.3) @	7.3 (6.3)	7.2 (7.2) @	12.1 (7.2)	1.4 (6.3) @	9.9 (6.3)	-16.5 (7.2) @	4.2 (7.2)
COPy Lag	54.0 (3.1) **	55.1 (3.1) **	56.8 (3.6) ** &	60.4 (3.6) ** &	34.1 (3.1) **	31.3 (3.1) **	31.0 (3.6) ** &	33.2 (3.6) ** &	44.2 (3.1) **	42.8 (3.1) **	52.6 (3.6) ** & #	63.7 (3.6) ** & #
COPy Lead	-39.2 (3.8) **	-37.8 (3.8) **	-33.3 (4.4) **	-35.1 (4.4) **	-27.3 (3.8) **	-26.3 (3.8) **	-26.0 (4.4) **	-26.0 (4.4) **	-45.7 (3.8) **	-37.4 (3.8) **	-48.4 (4.4) **	-42.4 (4.4) **

**: significantly different from the value in the other two frequency bands across all condition;
#: significantly different from the value for the females in same frequency band.
@: significantly different from the value in the eyes closed phase in all conditions
&: significantly different from the value for the females in all conditions
Significance level at $p<0.05$.

Table 4-3 Least square mean of the average value of the variables for the phase lock, phase lock + phase lag, phase lock + phase lead portion of the time series for elderly age group in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). The transfer function values (A/B) were calculated for the phase lock portion only. Data reported as Mean (SEM). (n = 14)

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
EMG/SBP	5.50 (0.64) E-05	5.57 (0.64) E-05	6.35 (0.74) E-05	6.56 (0.74) E-05	5.56 (0.64) E-05	5.53 (0.64) E-05	6.32 (0.74) E-05	6.54 (0.74) E-05	5.61 (0.64) E-05	5.53 (0.64) E-05	6.57 (0.74) E-05	6.67 (0.74) E-05
EMG/DBP	1.08 (0.15) E-04	1.10 (0.15) E-04	1.39 (0.18) E-04	1.46 (0.18) E-04	1.09 (0.15) E-04	1.10 (0.15) E-04	1.38 (0.18) E-04	1.46 (0.18) E-04	1.10 (0.15) E-04	1.11 (0.15) E-04	1.44 (0.18) E-04	1.53 (0.18) E-04
EMG/CO	1.24 (0.13) E-03	1.23 (0.13) E-03	1.17 (0.15) E-03	1.10 (0.15) E-03	1.25 (0.13) E-03	1.25 (0.13) E-03	1.15 (0.15) E-03	1.09 (0.15) E-03	1.26 (0.13) E-03	1.29 (0.13) E-03	1.18 (0.15) E-03	1.14 (0.15) E-03
EMG/COPx	9.62 (18.66)	37.49 (18.66)	-20.53 (21.55)	-34.63 (21.55)	10.83 (18.66)	18.68 (18.66)	-19.98 (21.55)	-18.12 (21.55)	12.44 (18.66)	-14.54 (18.66)	-20.83 (21.55)	-33.71 (21.55)
EMG/COPy	29.55 (28.32)	-11.86 (28.32)	-13.98 (32.70)	-16.47 (32.70)	73.85 (28.32)	-0.74 (28.32)	-13.66 (32.70)	-18.81 (32.70)	-18.04 (28.32)	0.25 (28.32)	-14.00 (32.70)	-17.11 (32.70)
EMG	7.31 (0.91) E-03	7.64 (0.91) E-03	9.08 (1.05) E-03	9.20 (1.05) E-03	7.37 (0.91) E-03	7.60 (0.91) E-03	9.03 (1.05) E-03	9.15 (1.05) E-03	7.47 (0.91) E-03	7.62 (0.91) E-03	9.44 (1.05) E-03	9.43 (1.05) E-03

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
SBP	133 (9) @	138 (9)	143 (10) @	146 (10)	133 (9) @	138 (9)	144 (10) @	146 (10)	133 (9) @	139 (9)	143 (10) @	147 (10)
EMG	7.29 (0.89) E-03	7.67 (0.89) E-03	8.96 (1.03) E-03	9.13 (1.03) E-03	7.37 (0.89) E-03	7.52 (0.89) E-03	9.00 (1.03) E-03	9.16 (1.03) E-03	7.42 (0.89) E-03	7.48 (0.89) E-03	9.35 (1.03) E-03	9.33 (1.03) E-03
SBP Lag	133 (9) @	138 (9)	143 (10) @	147 (10)	133 (9) @	138 (9)	145 (10) @	146 (10)	133 (9) @	139 (9)	143 (10) @	147 (10)
EMG	7.33 (0.94) E-03	7.63 (0.94) E-03	9.16 (1.09) E-03	9.26 (1.09) E-03	7.39 (0.94) E-03	7.62 (0.94) E-03	9.01 (1.09) E-03	9.15 (1.09) E-03	7.58 (0.94) E-03	7.63 (0.94) E-03	9.66 (1.09) E-03	9.33 (1.09) E-03
SBP Lead	133 (9) @	138 (9)	144 (10) @	146 (10)	133 (9) @	138 (9)	143 (10) @	146 (10)	134 (9) @	139 (9)	142 (10) @	147 (10)
EMG	7.29 (0.93) E-03	7.62 (0.93) E-03	9.10 (1.08) E-03	9.19 (1.08) E-03	7.39 (0.93) E-04	7.65 (0.93) E-03	9.08 (1.08) E-03	9.28 (1.08) E-03	7.46 (0.93) E-03	7.72 (0.93) E-03	9.43 (1.08) E-03	9.66 (1.08) E-03
DBP	69 (4)	70 (4)	67 (4)	66 (4)	69 (4)	71 (4)	67 (4)	66 (4)	69 (4)	71 (4)	67 (4)	66 (4)
EMG	7.31 (0.92) E-03	7.62 (0.92) E-03	9.06 (1.06) E-03	9.33 (1.06) E-03	7.35 (0.92) E-03	7.55 (0.92) E-03	9.16 (1.06) E-03	9.16 (1.06) E-03	7.44 (0.92) E-03	7.47 (0.92) E-03	9.36 (1.06) E-03	9.78 (1.06) E-03
DBP Lag	69 (4)	71 (4)	67 (4)	66 (4)	68 (4)	71 (4)	67 (4)	66 (4)	69 (4)	71 (4)	67 (4)	66 (4)

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
EMG	7.28 (0.97) E-03	7.61 (0.97) E-03	9.18 (1.12) E-03	9.07 (1.12) E-03	7.42 (0.97) E-03	7.74 (0.97) E-03	8.99 (1.12) E-03	9.40 (1.12) E-03	7.54 (0.97) E-03	7.97 (0.97) E-03	9.58 (1.12) E-03	9.53 (1.12) E-03
DBP Lead	69 (4)	70 (4)	67 (4)	66 (4)	69 (4)	71 (4)	67 (4)	65 (4)	69 (4)	71 (4)	66 (4)	66 (4)
EMG	7.29 (0.93) E-03	7.57 (0.93) E-03	9.23 (1.08) E-03	9.23 (1.08) E-03	7.40 (0.93) E-03	7.63 (0.93) E-03	9.10 (1.08) E-03	9.24 (1.08) E-03	7.45 (0.93) E-03	7.81 (0.93) E-03	9.47 (1.08) E-03	9.54 (1.08) E-03
CO	6.04 (0.57) @	6.24 (0.57)	8.23 (0.66) @ #	8.75 (0.66) #	6.05 (0.57) @	6.21 (0.57)	8.25 (0.66) @ #	8.79 (0.66) #	6.06 (0.57) @	6.18 (0.57)	8.30 (0.66) @ #	8.64 (0.66) #
EMG	7.28 (0.91) E-03	7.57 (0.91) E-03	9.10 (1.05) E-03	9.38 (1.05) E-03	7.36 (0.91) E-03	7.67 (0.91) E-03	9.05 (1.05) E-03	9.32 (1.05) E-03	7.58 (0.91) E-03	7.96 (0.91) E-03	9.07 (1.05) E-03	9.35 (1.05) E-03
CO Lag	6.03 (0.57) @	6.24 (0.57)	8.19 (0.65) @ #	8.69 (0.65) #	6.05 (0.57) @	6.21 (0.57)	8.27 (0.65) @ #	8.80 (0.65) #	6.03 (0.57) @	6.18 (0.57)	8.35 (0.65) @ #	8.64 (0.65) #
EMG	7.30 (0.96) E-03	7.55 (0.96) E-03	9.33 (1.11) E-03	9.11 (1.11) E-03	7.42 (0.96) E-03	7.59 (0.96) E-03	9.22 (1.11) E-03	9.21 (1.11) E-03	7.38 (0.96) E-03	7.63 (0.96) E-03	9.63 (1.11) E-03	9.85 (1.11) E-03
CO Lead	6.05 (0.57) @	6.23 (0.57)	8.24 (0.66) @ #	8.80 (0.66) #	6.05 (0.57) @	6.21 (0.57)	8.24 (0.66) @ #	8.79 (0.66) #	6.08 (0.57) @	6.17 (0.57)	8.29 (0.66) @ #	8.64 (0.66) #
EMG	7.34 (0.93) E-03	7.63 (0.93) E-03	9.06 (1.08) E-03	9.27 (1.08) E-03	7.31 (0.93) E-03	7.57 (0.93) E-03	9.03 (1.08) E-03	9.20 (1.08) E-03	7.41 (0.93) E-03	7.74 (0.93) E-03	9.50 (1.08) E-03	9.37 (1.08) E-03

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
COPx	5.32 (1.18) E-03 @	5.14 (1.18) E-03	2.60 (1.37) E-03 @	2.48 (1.37) E-03	5.33 (1.18) E-03 @	5.15 (1.18) E-03	2.61 (1.37) E-03 @	2.49 (1.37) E-03	5.33 (1.18) E-03 @	5.10 (1.18) E-03	2.66 (1.37) E-03 @	2.50 (1.37) E-03
EMG	7.35 (0.95) E-03	7.61 (0.95) E-03	9.09 (1.10) E-03	9.23 (1.10) E-03	7.34 (0.95) E-03	7.65 (0.95) E-03	8.94 (1.10) E-03	9.22 (1.10) E-03	7.49 (0.95) E-03	7.90 (0.95) E-03	9.63 (1.10) E-03	9.26 (1.10) E-03
COPx Lag	5.32 (1.18) E-03 @	5.14 (1.18) E-03	2.62 (1.37) E-03 @	2.45 (1.37) E-03	5.33 (1.18) E-03 @	5.17 (1.18) E-03	2.53 (1.37) E-03 @	2.49 (1.37) E-03	5.34 (1.18) E-03 @	5.08 (1.18) E-03	2.68 (1.37) E-03 @	2.53 (1.37) E-03
EMG	7.33 (0.87) E-03	7.63 (0.87) E-03	9.06 (1.01) E-03	9.27 (1.01) E-03	7.29 (0.87) E-03	7.52 (0.87) E-03	9.04 (1.01) E-03	9.16 (1.01) E-03	7.35 (0.87) E-03	7.39 (0.87) E-03	8.96 (1.01) E-03	9.34 (1.01) E-03
COPx Lead	5.32 (1.18) E-03 @	5.14 (1.18) E-03	2.60 (1.36) E-03 @	2.48 (1.36) E-03	5.33 (1.18) E-03 @	5.15 (1.18) E-03	2.62 (1.36) E-03 @	2.50 (1.36) E-03	5.32 (1.18) E-03 @	5.16 (1.18) E-03	2.61 (1.36) E-03 @	2.47 (1.36) E-03
EMG	7.32 (0.94) E-03	7.65 (0.94) E-03	9.07 (1.09) E-03	9.23 (1.09) E-03	7.31 (0.94) E-03	7.63 (0.94) E-03	9.21 (1.09) E-03	9.21 (1.09) E-03	7.42 (0.94) E-03	7.71 (0.94) E-03	9.41 (1.09) E-03	9.45 (1.09) E-03
COPy	2.27 (48.6) E-05	2.31 (48.6) E-05	-6.29 (5.61) E-04	-6.06 (5.61) E-04	0.84 (48.6) E-05	0.89 (48.6) E-05	-6.02 (5.61) E-04	-5.95 (5.61) E-04	2.49 (48.6) E-04	0.48 (48.6) E-05	-6.01 (5.61) E-04	-5.93 (5.61) E-04
EMG	7.31 (0.93) E-03	7.65 (0.93) E-03	9.04 (1.08) E-03	9.24 (1.08) E-03	7.28 (0.93) E-03	7.65 (0.93) E-03	9.20 (1.08) E-03	9.17 (1.08) E-03	7.54 (0.93) E-03	7.27 (0.93) E-03	9.25 (1.08) E-03	9.88 (1.08) E-03

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
COPy Lag	2.17 (47.9) E-05	2.19 (47.9) E-05	-6.34 (5.53) E-04	-6.10 (5.53) E-04	0.92 (47.9) E-05	-0.28 (47.9) E-05	-5.73 (5.53) E-04	-5.80 (5.53) E-04	7.65 (47.9) E-04	0.79 (47.9) E-05	-6.68 (5.53) E-04	-5.06 (5.53) E-04
EMG	7.37 (0.96) E-03	7.62 (0.96) E-03	9.20 (1.11) E-03	9.17 (1.11) E-03	7.34 (0.96) E-03	7.63 (0.96) E-03	9.22 (1.11) E-03	9.25 (1.11) E-03	7.30 (0.96) E-03	8.25 (0.96) E-03	9.64 (1.11) E-03	9.17 (1.11) E-03
COPy Lead	2.28 (49.2) E-05	1.97 (49.2) E-05	-6.24 (5.68) E-04	-5.93 (5.68) E-04	0.93 (49.2) E-05	4.17 (49.2) E-05	-6.56 (5.68) E-04	-6.25 (5.68) E-04	-1.50 (49.2) E-05	1.60 (49.2) E-05	-5.37 (5.68) E-04	-6.32 (5.68) E-04
@: significantly different from the value in the eyes closed phase in all conditions; #: significantly different from the value for females in all conditions Significance level at $p<0.05$												

Table 4-4 Least square mean for the average value of coherence > threshold in elderly participants in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). (n = 14). Data reported as Mean (SEM)

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
SBP	0.351 (0.013) #	0.339 (0.013) #	0.339 (0.016) #	0.298 (0.016) #	0.398 (0.013) #	0.394 (0.013) #	0.407 (0.016) #	0.403 (0.016) #	0.473 (0.013) #	0.448 (0.013) #	0.432 (0.016) #	0.447 (0.016) #
DBP	0.354 (0.015) #	0.347 (0.015) #	0.344 (0.017) #	0.322 (0.017) #	0.392 (0.015) #	0.401 (0.015) #	0.392 (0.017) #	0.400 (0.017) #	0.464 (0.015) #	0.439 (0.015) #	0.471 (0.017) #	0.448 (0.017) #
CO	0.338 (0.012) #	0.332 (0.012) #	0.304 (0.013) #	0.293 (0.013) #	0.406 (0.012) #	0.403 (0.012) #	0.406 (0.013) #	0.417 (0.013) #	0.451 (0.012) #	0.443 (0.012) #	0.451 (0.013) #	0.426 (0.013) #
COPx	0.412 (0.023) @	0.432 (0.023) @	0.480 (0.027)	0.506 (0.027)	0.429 (0.023) @	0.434 (0.023) @	0.454 (0.027)	0.470 (0.027)	0.481 (0.023) @ #	0.489 (0.023) @ #	0.525 (0.027) #	0.561 (0.027) #
COPy	0.417 (0.021)	0.420 (0.021)	0.444 (0.024)	0.490 (0.024)	0.433 (0.021)	0.434 (0.021)	0.439 (0.024)	0.460 (0.024)	0.449 (0.021) #	0.455 (0.021) #	0.485 (0.024) #	0.510 (0.024) #

#: significantly different from the values in the other two frequency bands;
@: significantly different from the value for the males in all conditions;
Significance level at $p<0.05$

Table 4-5 Least square mean of the average value of variables in the time periods where the coherence > threshold for the elderly participants in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz) (n = 14).Data reported as Mean (SEM)

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
EMG	7.32 (0.93) E-03	7.64 (0.93) E-03	9.05 (1.07) E-03	9.20 (1.07) E-03	7.34 (0.93) E-03	7.62 (0.93) E-03	9.23 (1.07) E-03	9.31 (1.07) E-03	7.51 (0.93) E-03	7.77 (0.93) E-03	9.51 (1.07) E-03	1.00 (0.10) E-02
SBP	133 (9) #	138 (9)	143 (10) #	146 (10)	133 (9) #	139 (9)	143 (10) #	146 (10)	133 (9) #	139 (9)	143 (10) #	147 (10)
EMG 2	7.32 (0.95) E-03	7.62 (0.95) E-03	9.06 (1.11) E-03	9.22 (1.11) E-03	7.36 (0.95) E-03	7.57 (0.95) E-03	9.18 (1.11) E-03	9.33 (1.11) E-03	7.65 (0.95) E-03	8.01 (0.95) E-03	9.50 (1.11) E-03	9.70 (1.11) E-03
DBP	69 (4)	70 (4)	67 (4)	66 (4)	69 (4)	71 (4)	67 (4)	66 (4)	69 (4)	71 (4)	67 (4)	66 (4)
EMG 3	7.31 (0.92) E-03	7.63 (0.92) E-03	9.06 (1.06) E-03	9.24 (1.06) E-03	7.39 (0.92) E-03	7.41 (0.92) E-03	9.00 (1.06) E-03	9.33 (1.06) E-03	7.53 (0.92) E-03	7.69 (0.92) E-03	9.45 (1.06) E-03	9.44 (1.06) E-03

	LF				VLF				ULF			
	F		M		F		M		F		M	
	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC	EO	EC
CO	6.05 (0.57) # @	6.21 (0.57) @	8.22 (0.66) #	8.72 (0.66)	6.05 (0.57) # @	6.21 (0.57) @	8.23 (0.66) #	8.79 (0.66)	6.06 (0.57) # @	6.19 (0.57) @	8.29 (0.66 #)	8.61 (0.66)
EMG 4	7.32 (0.95) E-03	7.63 (0.95) E-03	9.07 (1.10) E-03	9.23 (1.10) E-03	7.36 (0.95) E-03	7.53 (0.95) E-03	9.09 (1.10) E-03	9.37 (1.10) E-03	7.50 (0.95) E-03	7.73 (0.95) E-03	9.65 (1.10) E-03	9.40 (1.10) E-03
COP x	5.33 (1.18) E-03 #	5.14 (1.18) E-03	2.61 (1.37) E-03 #	2.48 (1.37) E-03	5.33 (1.18) E-03 #	5.16 (1.18) E-03	2.61 (1.37) E-03 #	2.48 (1.37) E-03	5.28 (1.18) E-03 #	5.11 (1.18) E-03	2.65 (1.37) E-03 #	2.49 (1.37) E-03
EMG 5	7.31 (0.91) E-03	7.62 (0.91) E-03	9.05 (1.06) E-03	9.23 (1.06) E-03	7.33 (0.91) E-04	7.50 (0.91) E-03	9.08 (1.06) E-03	9.29 (1.06) E-03	7.64 (0.91) E-03	7.96 (0.91) E-03	9.50 (1.06) E-03	9.59 (1.06) E-03
COP y	2.38 (48.3) E-05	1.69 (48.3) E-05	-6.25 (5.57) E-04	-6.04 (5.57) E-04	2.99 (48.3) E-04	0.22 (48.3) E-05	-6.59 (5.57) E-04	-5.97 (5.57) E-04	7.39 (48.3) E-05	-4.94 (48.3) E-05	-6.13 (5.57) E-04	-6.81 (5.57) E-04

#: significantly different from the value in the eyes closed condition;
@: significantly different from the value for males in all frequency bands;
Significance level at $p<0.05$

Table 4-6 Least square mean values of the SDA output for comparison in the elderly participant group for gender and eyes condition (n = 14). Data reported as Mean (SEM)

	M		F	
	EO	EC	EO	EC
delta txc	0.707(0.038)	0.675(0.038)	0.669(0.034)	0.699(0.035)
X2	1.83 (73.7) E-06	1.50 (73.7)E-06	1.20 (63.8)E-06	1.60 (63.9)E-06
Dxs	3.67 (47.4)E-06	2.83 (47.4)E-06	2.10 (41.1)E-06	3.30 (41.1)E-06
Dxl	4.02E-21(4.52E-05)	2.75E-21(4.52E-05)	-4.02E-21(3.92E-05)	-2.75E-21(3.92E-05)
Hxs	0.118(0.071)	0.132(0.071)	0.089(0.063)	0.169(0.065)
Hxl	0.103(0.056) #	0.079(0.056)	0.194(0.049) #	0.155(0.049)
delta tyc	0.708(0.038)	0.673(0.038)	0.671(0.034)	0.696(0.035)
Y2	1.83 (74.3)E-06	1.50 (74.3)E-06	1.20 (64.4)E-06	1.60 (64.4)E-06
Dys	3.50 (47.2)E-06	2.83 (47.2)E-06	1.98 (40.9)E-06	3.17 (41.0)E-06
Dyl	4.02E-21(4.52E-05)	2.75E-21(4.52E-05)	-4.02E-21(3.92E-05)	-2.75E-21(3.92E-05)

	M		F	
	EO	EC	EO	EC
Hys	0.122(0.071)	0.133(0.071)	0.080(0.063)	0.171(0.065)
Hyl	0.104(0.056) #	0.079(0.056)	0.194(0.049) #	0.155(0.049)
Delta trc	0.708(0.037)	0.675(0.037)	0.669(0.033)	0.699(0.035)
R2	0.40 (14.8)E-05	0.30 (14.8)E-05	0.23 (12.8)E-05	0.33 (12.8)E-05
Drs	7.33 (94.5)E-06	5.83(94.5)E-06	4.38 (82.0)E-06	6.37 (82.0)E-06
Drl	6.04E-21(9.04E-05)	7.52E-21(9.04E-05)	0.15 (78.3)E-06	-0.05 (78.3)E-06
Hrs	0.161(0.069)	0.176(0.069)	0.132(0.061)	0.211(0.063)
Hrl	0.104(0.056) #	0.079(0.056)	0.194(0.049) #	0.155(0.049)
#: significantly different from the value in the eyes closed condition; Significance level at $p<0.05$				

Table 4-7 Percentage of time of significance coherence over total 4 minute duration in elderly. (n = 14), eyes closed condition in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz).

		MEDIAN		MIN		MAX
SBP	LF	57.8		31.5		74.6
	VLF	42.1		17.8		56.8
	ULF	47.0		3.5		78.5
DBP	LF	60.9		33.3		78.3
	VLF	39.8		33.1		54.2
	ULF	49.4		2.4		83.5
CO	LF	61.4		30.8		72.4
	VLF	42.8		28.5		69.8
	ULF	48.0		22.3		70.3
COPx	LF	86.3		73.6		97.7
	VLF	60.4		30.2		89.3
	ULF	58.0		49.3		99.7
COPy	LF	89.1		75.0		98.1
	VLF	68.1		42.0		83.1
	ULF	56.9		23.0		99.7

4.4.2 Comparison between the young and the old age group

4.4.2.1 SDA analysis

The scaling exponent in both short-term and long-term regions showed significant difference with change in eyes condition (Table 4-8). The short-term exponent increased with eyes closed in all three directions (Hxs: eo: 0.098 ± 0.03, ec: 0.153 ± 0.03; Hys: eo 0.093 ± 0.03, ec 0.152 ± 0.03; Hrs: eo: 0.139 ± 0.03, ec: 0.193 ± 0.03). On the contrary the long-term scaling exponent showed a reduction with eyes closed in all three directions (Hxl: eo: 0.163 ± 0.02, ec: 0.140 ± 0.02; Hyl: eo: 0.163 ± 0.02, ec: 0.140 ± 0.02; Hrl: eo: 0.163 ± 0.02, ec: 0.140 ± 0.02). The transition time from short-term to long-term strategy was significantly lower in elderly age group (delta txc: old: 0.687 ± 0.02, young 0.762 ± 0.02; delta tyc: old: 0.687 ± 0.02, young 0.762 ± 0.02; delta trc: old 0.687 ± 0.02, young 0.761 ± 0.02).

4.4.2.2 Cardiovascular-postural system relationship

The transfer value for EMG to CO was significantly different and highest in ULF band ((1.33 ± 0.0) E-03, $p<0.05$) (Table 4-9). The mean magnitude for EMG in both phase lock lag and lead was significantly higher in the older ((8.43 ± 0.6) E-03) group than in the younger ((6.3 ± 0.6) E-03) group. The mean value of SBP in phase lock lag and lead was significantly higher in the older (142 ± 5) than in the younger age group (114 ± 5). The average value of SBP in phase lock was significantly higher in ULF (129 ± 4); and, in phase lag it was significantly higher in ULF (129 ± 4). When DBP was tested for association with EMG, the average

value of EMG during DBP phase lag was significantly higher in males ((8.3 ± 0.6) E-03) than females ((6.4 ± 0.6) E-03).

The average value of EMG in phase lock with CO time period, was significantly higher in ULF ((7.5 ± 0.4) E-03) than LF ((7.4 ± 0.4) E-03). In the elderly the average value of EMG was significantly higher in ULF ((8.6 ± 0.6) E-03) than VLF ((8.4 ± 0.6) E-03) and LF ((8.3 ± 0.6) E-03). The average value for CO in both phase lock lag and lead showed a significantly higher value in older (7.46 ± 0.4) compared to the younger (4.43 ± 0.4) age group; and, significantly high value in males (6.68 ± 0.4) versus females (5.22 ± 0.4) . Additionally CO was significant lower in ULF (5.91 ± 0.2) than the other two bands (5.97 ± 0.2) for phase lock and significantly lower in ULF (5.90 ± 0.2) versus VLF (5.97 ± 0.2) with phase lag.

The average value for the COPx was significantly lower in young ((-4.8 ± 2.9) E-03) compared to old ((3.8 ± 2.8) E-03) in both phase lock lag and lead. The average value for EMG with COPy lag showed a significant difference between males ((8.3 ± 0.6) E-03) and females ((6.4 ± 0.6) E-03). Average value of COPy did not show any association with any conditions.

The average phase angle of SBP in phase lag was significantly lower in VLF (31.8 ± 2.1) than the other two bands (LF: 39.9± 2.1; ULF: 41.1 ± 2.1) and in phase lead was significantly lower in VLF (-34.7 ± 2.1) than ULF (-44.1 ± 2.1) and LF (-39.5 ± 2.1) (Table 4-10). Similarly, the average phase angle of DBP in phase lag was significantly lower in VLF (33.3 ± 2.1) than LF (41.9 ± 2.1) and ULF (45.6 ± 2.1) bands and, in the case of phase lead, was significantly lower in

VLF (-34.3 ± 2.3) than LF (-43.4 ± 2.3) and ULF (-42.4 ± 2.3). The average phase angle of CO in phase lag case was significantly different in all three bands LF (39.8 ± 1.9), VLF (33.9 ± 1.9) and ULF (47.4 ± 1.9) and in phase lead case it ws significantly high in ULF (-45.9 ± 2.4) than LF (-37.5 ± 2.4) and VLF (-32.8 ± 2.4).

The average phase angle of COPx in the phase lock condition was significantly different in all three frequency bands LF (-30.2 ± 4.3), VLF (-14.3 ± 4.3) and ULF (3.5 ± 4.3); significant difference was found between young males (-1.2 ± 6.8) and females (-25.5 ± 7.3). The average phase angle for COPx in the lag phase was significantly lower in the older (33.9 ± 2.0) versus younger (39.9 ± 2.0) and significantly lower in VLF (29.4 ± 2.81) than both ULF (41.9 ± 2.81) and LF (39.4 ± 2.81) frequency bands. Similarly, the average phase angle with phase lead was significantly lower in LF (-60.2 ± 2.7) than VLF (-38.1 ± 2.7) and ULF (-36.6 ± 2.7); and young females (-53.2 ± 4.3) had significantly higher values than young males (-38.2 ± 4.0).

The average phase angle for the COPy in phase lock was significantly higher in LF (28.8 ±3.8) than the VLF (8.5 ± 3.8) and ULF (3.3 ± 3.8) bands. With phase lag, the value for VLF (32.6 ± 1.8) was significantly lower than the other (54.3 ± 1.8) bands; and, older males (52.4 ± 2.78) has significantly higher values than younger males (43.1 ± 2.6) and older females (43.1 ± 2.4). For the phase lead condition, the COPy values were significantly different between all three bands: LF (-36.6 ± 2.4), VLF (-26.5 ± 2.4) and ULF (-43.9 ± 2.4).

The average value for coherence was significantly different in all three frequency bands for all combinations of EMG – SBP (LF: 0.32 ± 0.1; VLF: 0.40 ±

0.1; ULF: 0.44 ± 0.1), EMG – DBP (LF: 0.33 ± 0.1; VLF: 0.40 ± 0.1; ULF: 0.44 ± 0.1), EMG – CO (LF: 0.32 ± 0.1; VLF: 0.40 ± 0.1; ULF: 0.44 ± 0.1) (Table 4-11). Both EMG - COPx (LF: 0.47 ± 0.01; VLF: 0.47 ± 0.01; ULF: 0.53 ± 0.01) and EMG - COPy (LF: 0.44 ± 0.01; VLF: 0.45 ± 0.01; ULF: 0.48 ± 0.01) had significantly higher coherence in ULF than the LF and VLF bands. EMG – COPx had higher coherence in males (0.51 ± 0.01) than females (0.47 ± 0.01). Males in the elderly group had higher coherence (0.49 ± 0.01) for the EMG – COPy combination than young males (0.44 ± 0.01) and elderly females (0.43 ± 0.01).

The average value of EMG for the time segments with significance coherence (Table 4-12) was significantly higher in the elderly ((8.5 ± 0.6) E-03) than the young ((6.5 ± 0.6) E-03) age group. The average value of EMG for the EMG – SBP and EMG – CO combinations was statistically higher in males (CO: (8.3 ± 0.1) E-03; SBP: (8.3 ± 0.1) E-03) than females (CO: (6.4 ± 0.1) E-03; SBP: (6.5 ± 0.1) E-03). The average value for SBP was significantly higher in old (142 ± 5) versus young (114 ± 5). The average value for CO was significantly higher in old (7.45 ± 0.37) versus young (4.43 ± 0.38); and, significantly higher in males (6.67 ± 0.38) than females (5.22 ± -.37). COPx was significantly higher in the older ((3.8 ± 2.8) E-03) compared younger ((-4.8 ± 2.9) E-03) age group. Finally, COPy was significantly lower in the ULF ((15.8 ± 16.7) E-03) than the LF and VLF bands ((15.9 ± 16.7) E-03).

Table 4-8 Least square mean values for the SDA output for the young and elderly participants in the eyes closed/open phase. Comparison is across age, gender and eyes condition (young n = 13; old n = 14). Data reported as Mean (SEM).

	YOUNG				OLD			
	M		F		M		F	
	EO	EC	EO	EC	EO	EC	EO	EC
delta txc	0.807(0.032) #	0.811(0.033) #	0.713(0.032) #	0.720(0.029) #	0.707(0.038)	0.675(0.038)	0.669(0.034)	0.699(0.035)
X2	1.28 (0.6) E-04	1.38 (0.6)E-04	0.9(57.1)E-06	0.9 (57.1)E-06	1.83 (73.7) E-06	1.50 (73.7)E-06	1.20 (63.8)E-06	1.60 (63.9)E-06
Dxs	8.04 (3.87)E-05	9.42 (3.88)E-05	1.41 (36.8)E-06	1.70 (36.7)E-06	3.67 (47.4)E-06	2.83 (47.4)E-06	2.10 (41.1)E-06	3.30 (41.1)E-06
Dxl	7.14 (3.69)E-05	7.41 (3.69)E-05	4.02E-21(3.51E-05)	2.75E-21(3.50E-05)	4.02E-21(4.52E-05)	2.75E-21(4.52E-05)	-4.02E-21(3.92E-05)	-2.75E-21(3.92E-05)
Hxs	0.210(0.059) &	0.229(0.060)	-0.023(0.058) &	0.082(0.055)	0.118(0.071) &	0.132(0.071)	0.089(0.063) &	0.169(0.065)
Hxl	0.242(0.046) &	0.222(0.046)	0.116(0.044) &	0.106(0.043)	0.103(0.056) &	0.079(0.056)	0.194(0.049) &	0.155(0.049)
delta tyc	0.811(0.032) #	0.810(0.033) #	0.712(0.032) #	0.716(0.029) #	0.708(0.038)	0.673(0.038)	0.671(0.034)	0.696(0.035)
Y2	1.30 (0.6) E-04	1.38 (0.6) E-04	0.86 (57.6)E-06	1.00 (57.6)E-06	1.83 (74.3)E-06	1.50 (74.3)E-06	1.20 (64.4)E-06	1.60 (64.4)E-06
Dys	8.02 (3.86)E-05	9.41 (3.86)E-05	1.51 (36.7)E-06	1.80 (36.6)E-06	3.50 (47.2)E-06	2.83 (47.2)E-06	1.98 (40.9)E-06	3.17 (41.0)E-06

	YOUNG				OLD			
	M		F		M		F	
	EO	EC	EO	EC	EO	EC	EO	EC
Dyl	7.14 (3.69)E-05	7.41 (3.69)E-05	4.02E-21(3.51E-05)	2.75E-21(3.50E-05)	4.02E-21(4.52E-05)	2.75E-21(4.52E-05)	-4.02E-21(3.92E-05)	-2.75E-21(3.92E-05)
Hys	0.197(0.059) &	0.227(0.061)	-0.025(0.059) &	0.081(0.055)	0.122(0.071) &	0.133(0.071)	0.080(0.063) &	0.171(0.065)
Hyl	0.241(0.046) &	0.222(0.046)	0.115(0.044) &	0.107(0.043)	0.104(0.056) &	0.079(0.056)	0.194(0.049) &	0.155(0.049)
Delta trc	0.809(0.031) #	0.809(0.032) #	0.712(0.032) #	0.716(0.029) #	0.708(0.037)	0.675(0.037)	0.669(0.033)	0.699(0.035)
R2	2.57 (1.21)E-04	2.73 (1.21)E-04	0.15 (11.4)E-05	0.20 (11.4)E-05	0.40 (14.8)E-05	0.30 (14.8)E-05	0.23 (12.8)E-05	0.33 (12.8)E-05
Drs	1.61 (0.8) E-04	1.88 (0.8) E-04	2.48 (73.4)E-06	3.20 (73.2)E-06	7.33 (94.5)E-06	5.83(94.5)E-06	4.38 (82.0)E-06	6.37 (82.0)E-06
Drl	1.42 (0.7)E-04	1.48 (0.7)E-04	1.96E-20(7.01E-05)	2.11E-20(7.00E-05)	6.04E-21(9.04E-05)	7.52E-21(9.04E-05)	0.15 (78.3)E-06	-0.05 (78.3)E-06
Hrs	0.238(0.057) &	0.264(0.059)	0.026(0.057) &	0.124(0.053)	0.161(0.069) &	0.176(0.069)	0.132(0.061) &	0.211(0.063)
Hrl	0.242(0.046) &	0.221(0.046)	0.116(0.044) &	0.106(0.043)	0.104(0.056) &	0.079(0.056)	0.194(0.049) &	0.155(0.049)

: significantly different from values for old in all conditions
&: significantly different from value in the eyes closed phase in all conditions
Significance level at $p<0.05$

Table 4-9 Least Square mean value of the variables in the phase lock, phase lock + phase lag, phase lock + phase lead portion of the time series. The transfer function output (A/B) is calculated for the phase lock portion only. The comparison is in the eyes closed phase between gender and age in all three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). (young n = 13; old n = 14). Data reported as Mean (SEM).

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
EMG/ SBP	4.93 (0.8) E-05	6.10 (0.7) E-05	5.57 (0.7) E-05	6.56 (0.8) E-05	4.95 (0.8) E-05	6.08 (0.7) E-05	5.53 (0.7) E-05	6.54 (0.8) E-05	4.98 (0.8) E-05	5.91 (0.7) E-05	5.53 (0.7) E-05	6.67 (0.8) E-05
EMG/ DBP	8.30 (1.55) E-05	1.03 (0.14) E-04	1.10 (0.13) E-04	1.46 (0.15) E-04	8.26 (1.55) E-05	1.03 (0.14) E-04	1.10 (0.13) E-04	1.46 (0.15) E-04	8.22 (1.55) E-05	1.01 (0.14) E-05	1.11 (0.13) E-04	1.53 (0.15) E-04
EMG/ CO	1.27 (0.18) E-03	1.62 (0.17) E-03	1.23 (0.16) E-03	1.10 (0.18) E-03	1.28 (0.18) E-03	1.63 (0.17) E-03	1.25 (0.16) E-03	1.09 (0.18) E-03	1.30 (0.18) E-03	1.62 (0.17) E-03	1.29 (0.16) E-03	1.14 (0.18) E-03
EMG/ COPx	-4.39 (20.44)	6.35 (18.92)	37.49 (17.70)	-34.63 (20.44)	-3.32 (20.44)	6.98 (18.92)	18.68 (17.70)	-18.13 (20.44)	-4.60 (20.44)	5.77 (18.92)	-14.54 (17.70)	-33.71 (20.44)

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
EMG/ COPy	1.48 (13.95)	-0.64 (12.92)	-11.87 (12.08)	-16.48 (13.95)	1.54 (13.95)	5.74 (12.92)	-0.75 (12.08)	-18.82 (13.95)	1.59 (13.95)	-21.84 (12.92)	0.26 (12.08)	-17.11 (13.95)
EMG	5.39 (0.94) E-03 @	7.44 (0.87) E-03 @	7.64 (0.82) E-03	9.20 (0.94) E-03	5.37 (0.94) E-03@	7.40 (0.87) E-03 @	7.60 (0.82) E-03	9.15 (0.94) E-03	5.39 (0.94) E-03 @	7.21 (0.87) E-03 @	7.62 (0.82) E-03	9.43 (0.94) E-03
SBP	109 (8) @	120 (7) @	138 (7)	146 (8)	108 (8) @	120 (7) @	138 (7)	146 (8)	109 (8) @ &	121 (7) @ &	139 (7) &	147 (8) &
EMG	5.35 (0.93) E-03 @	7.34 (0.86) E-03 @	7.67 (0.80) E-03	9.13 (0.92) E-03	5.27 (0.92) E-03@	7.38 (0.86) E-03@	7.52 (0.80) E-03	9.16 (0.92) E-03	5.38 (0.92) E-03 @	6.97 (0.86) E-03 @	7.48 (0.80) E-03	9.33 (0.92) E-03
SBP Lag	109 (8) @	120 (7) @	138 (7)	147 (8)	108 (8) @	120 (7) @	138 (7)	146 (8)	109 (8) @ &	121 (7) @ &	139 (7) &	147 (8) &
EMG	5.41 (0.96) E-03 @	7.49 (0.89) E-03 @	7.63 (0.83) E-03	9.26 (0.96) E-03	5.47 (0.96) E-03 @	7.28 (0.89) E-03 @	7.62 (0.83) E-03	9.15 (0.96) E-03	5.40 (0.96) E-03 @	7.54 (0.89) E-03 @	7.63 (0.83) E-03	9.33 (0.96) E-03
SBP Lead	109 (8) @	120 (7) @	138 (7)	146 (8)	109 (8) @	120 (7) @	138 (7)	146 (8)	109 (8) @	120 (7) @	139 (7)	147 (8)

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
EMG	5.41 (0.94) E-03 @	7.31 (0.87) E-03 @	7.62 (0.81) E-03	9.19 (0.94) E-03	5.37 (0.94) E-03 @	7.39 (0.87) E-03 @	7.65 (0.82) E-03	9.28 (0.94) E-03	5.36 (0.94) E-03 @	7.22 (0.87) E-03 @	7.72 (0.82) E-03	9.66 (0.94) E-03
DBP	66 (4)	71 (4)	70 (4)	66 (4)	66 (4)	71 (4)	71 (4)	66 (4)	66 (4)	71 (4)	71 (4)	66 (4)
EMG	5.40 (0.93) E-03 @	7.38 (0.86) E-03 @ #	7.62 (0.80) E-03	9.33 (0.93) E-03 #	5.31 (0.93) E-03 @	7.40 (0.86) E-03 @ #	7.55 (0.81) E-03	9.16 (0.93) E-03 #	5.28 (0.93) E-03 @	7.07 (0.86) E-03 @ #	7.47 (0.81) E-03	9.78 (0.93) E-03 #
DBP Lag	66 (4)	71 (4)	71 (4)	66 (4)	66 (4)	71 (4)	71 (4)	66 (4)	66 (4)	71 (4)	71 (4)	66 (4)
EMG	5.43 (0.98) E-03 @	7.16 (0.91) E-03 @	7.61 (0.85) E-03	9.07 (0.98) E-03	5.46 (0.98) E-03 @	7.29 (0.90) E-03 @	7.74 (0.85) E-03	9.40 (0.98) E-03	5.42 (0.98) E-03 @	7.33 (0.91) E-03 @	7.97 (0.85) E-03	9.53 (0.98) E-03
DBP Lead	65 (4)	71 (4)	70 (4)	66 (4)	66 (4)	71 (4)	71 (4)	65 (4)	66 (4)	71 (4)	71 (4)	66 (4)
EMG	5.37 (0.94) E-03 @	7.34 (0.87) E-03 @	7.57 (0.81) E-03	9.23 (0.94) E-03	5.37 (0.94) E-03 @ &	7.35 (0.87) E-03 @ &	7.63 (0.81) E-03 &	9.24 (0.94) E-03 &	5.39 (0.94) E-03 @	7.24 (0.87) E-03 @	7.81 (0.81) E-03	9.54 (0.94) E-03

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
CO	4.25 (0.55) @	4.65 (0.51) @ #	6.24 (0.48)	8.75 (0.55) #	4.25 (0.55) @	4.63 (0.51) @ #	6.21 (0.48)	8.79 (0.55) #	4.19 (0.55) @ &	4.63 (0.51) @ # &	6.18 (0.48) &	8.64 (0.55) # &
EMG	5.39 (0.97) E-03 @	7.43 (0.90) E-03 @	7.57 (0.84) E-03	9.38 (0.97) E-03	5.37 (0.97) E-03 @	7.47 (0.90) E-03 @	7.67 (0.84) E-03	9.32 (0.97) E-03	5.43 (0.97) E-03 @	7.32 (0.90) E-03 @	7.96 (0.84) E-03	9.35 (0.97) E-03
CO Lag	4.26 (0.55) @ &	4.67 (0.51) @ # &	6.24 (0.47) &	8.69 (0.55) # &	4.25 (0.55) @	4.64 (0.51) @ #	6.21 (0.47)	8.80 (0.55) #	4.14 (0.55) @	4.63 (0.51) @ #	6.18 (0.47)	8.64 (0.55) #
EMG	5.37 (0.94) E-03 @	7.27 (0.87) E-03 @ #	7.55 (0.81) E-03	9.11 (0.94) E-03 #	5.35 (0.94) E-03 @	7.26 (0.87) E-03 @ #	7.59 (0.81) E-03	9.21 (0.94) E-03 #	5.48 (0.94) E-03 @	7.18 (0.87) E-03 @ #	7.63 (0.81) E-03	9.85 (0.94) E-03 #
CO Lead	4.25 (0.56) @	4.63 (0.51) @ #	6.23 (0.48)	8.80 (0.56) #	4.25 (0.56) @	4.63 (0.51) @ #	6.21 (0.48)	8.79 (0.56) #	4.24 (0.56) @	4.63 (0.51) @ #	6.17 (0.48)	8.64 (0.56) #
EMG	5.36 (0.94) E-03 @	7.32 (0.87) E-03 @	7.63 (0.82) E-03	9.27 (0.95) E-03	5.44 (0.95) E-03 @	7.36 (0.87) E-03 @	7.57 (0.82) E-03	9.20 (0.94) E-03	5.39 (0.94) E-03 @	7.32 (0.87) E-03 @	7.74 (0.82) E-03	9.37 (0.94) E-03

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
COPx	2.31 (4.37) E-03 @	-1.22 (0.40) E-02 @	5.14 (3.79) E-03	2.48 (4.37) E-03	2.30 (4.37) E-03 @	-1.19 (0.40) E-02 @	5.15 (3.79) E-03	2.49 (4.37) E-03	2.29 (4.37) E-03 @	-1.18 (0.40) E-02 @	5.10 (3.79) E-03	2.50 (4.37) E-03
EMG	5.41 (0.95) E-03 @	7.31 (0.89) E-03 @	7.61 (0.82) E-03	9.23 (0.95) E-03	5.39 (0.95) E-03 @	7.40 (0.88) E-03 @	7.65 (0.82) E-03	9.22 (0.95) E-03	5.40 (0.95) E-03 @	7.25 (0.88) E-03 @	7.90 (0.82) E-03	9.26 (0.95) E-03
COPx Lag	2.31 (4.34) E-03 @	-1.22 (0.40) E-02 @	5.14 (3.76) E-03	2.45 (4.34) E-03	2.32 (4.34) E-03 @	-1.19 (0.40) E-02 @	5.17 (3.76) E-03	2.49 (4.34) E-03	2.28 (4.34) E-03 @	-1.16 (0.40) E-02 @	5.08 (3.76) E-03	2.53 (4.34) E-03
EMG	5.35 (0.93) E-03 @	7.27 (0.86) E-03 @ #	7.63 (0.80) E-03	9.27 (0.93) E-03	5.45 (0.93) E-03 @	7.33 (0.86) E-03 @ #	7.52 (0.80) E-03	9.16 (0.93) E-03	5.44 (0.93) E-03 @	7.45 (0.86) E-03 @ #	7.39 (0.80) E-03	9.34 (0.93) E-03
COPx Lead	2.31 (4.43) E-03 @	-1.21 (0.41) E-02 @	5.14 (3.84) E-03	2.48 (4.43) E-03	2.29 (4.43) E-03 @	-1.21 (0.41) E-02 @	5.15 (3.84) E-03	2.50 (4.43) E-03	2.29 (4.43) E-03 @	-1.22 (0.41) E-02 @	5.16 (3.84) E-03	2.47 (4.43) E-03
EMG	5.37 (0.94) E-03 @	7.28 (0.87) E-03 @	7.65 (0.81) E-03	9.23 (0.94) E-03	5.40 (0.94) E-03 @	7.34 (0.87) E-03 @	7.63 (0.82) E-03	9.21 (0.94) E-03	5.38 (0.94) E-03 @	7.32 (0.87) E-03 @	7.71 (0.81) E-03	9.45 (0.94) E-03

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
COPy	0.41 (35.3) E-03	6.40 (3.27) E-02	2.31E-05 (3.06E-02)	-0.60 (35.3) E-03	0.39 (35.3) E-03	6.40 (3.27) E-02	8.90E-06 (3.06E-02)	-0.59 (35.3) E-03	0.36 (35.3) E-03	6.38 (3.27) E-02	4.80E-06 (3.06E-02	-0.59 (35.3) E-03
EMG	5.36 (0.96) E-03 @	7.30 (0.89) E-03 @ #	7.65 (0.83) E-03	9.24 (0.96) E-03	5.36 (0.96) E-03 @	7.16 (0.89) E-03 @ #	7.65 (0.83) E-03	9.17 (0.96) E-03	5.23 (0.96) E-03 @	7.30 (0.89) E-03 @ #	7.27 (0.83) E-03	9.88 (0.96) E-03
COPy Lag	0.41 (35.3) E-03	6.40 (3.27) E-02	2.19E-05 (3.06E-02	-0.61 (35.3) E-03	0.39 (35.3) E-03	6.36 (3.27) E-02	-2.80E-06 (3.06E-02	-0.58 (35.3) E-03	0.34 (35.3) E-03	6.42 (3.27) E-02	7.90E-06 (3.06E-02	-0.50 (35.3) E-03
EMG	5.44 (0.95) E-03 @	7.22 (0.88) E-03 @	7.62 (0.82) E-03	9.17 (0.95) E-03	5.43 (0.95) E-03 @	7.63 (0.88) E-03 @	7.63 (0.82) E-03	9.25 (0.95) E-03	5.55 (0.95) E-03 @	7.35 (0.88) E-03 @	8.25 (0.82) E-03	9.17 (0.95) E-03
COPy Lead	0.41 (35.3) E-03	6.40 (3.27) E-02	1.97E-05 (3.06E-02	-0.59 (35.3) E-03	0.40 (35.3) E-03	6.40 (3.27) E-02	4.17E-05 (3.06E-02	-0.62 (35.3) E-03	0.40 (35.3) E-03	6.37 (3.27) E-02	1.60E-05 (3.06E-02	-0.63 (35.3) E-03

&: significantly different from the values in the other two frequency bands in all conditions;
@: significantly different from the values for old in all conditions.
#: significantly different from the value for females in all condition
Significance level at $p < 0.05$

Table 4-10 Least square value of the phase angle between EMG and the corresponding variable for the phase lock, phase lock + phase lag, and phase lock +phase lead portions of the whole time series. Comparison in three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz) between young/old age group and male/female gender. (young n = 13; old n = 14). Data reported as Mean (SEM).

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
SBP	2.2 (6.4)	3.7 (6.0)	-2.6 (5.6)	0.1 (6.4)	-4.9 (6.4)	-1.1 (6.0)	-4.9 (5.6)	-5.5 (6.4)	-5.5 (6.4)	-3.2 (6.0)	10.0 (5.6)	6.8 (6.4)
SBP Lag	41.1 (4.5)	36.3 (4.2)	40.9 (3.9)	41.6 (4.5)	29.9 (4.5) &	30.6 (4.2) &	32.0 (3.9) &	34.6 (4.5) &	40.7 (4.5)	33.0 (4.2)	46.4 (3.9)	44.3 (4.5)
SBP Lead	-39.2 (4.3)	-37.2 (4.0)	-39.7 (3.8)	-41.8 (4.3)	-37.4 (4.3) &	-35.7 (4.0) &	-32.0 (3.8) &	-33.7 (4.3) &	-49.1 (4.3)	-41.7 (4.0)	-41.2 (3.8)	-44.3 (4.3)
DBP	4.1 (6.0)	9.9 (5.6)	-5.4 (5.2)	4.1 (6.0)	0.8 (6.0)	-2.2 (5.6)	-5.1 (5.2)	0.3 (6.0)	-4.4 (6.0)	0.1 (5.6)	11.4 (5.2)	2.4 (6.0)
DBP Lag	41.4 (4.3)	40.8 (4.0)	38.6 (3.8)	47.1 (4.3)	34.3 (4.3) &	36.3 (4.0) &	28.7 (3.8) &	34.0 (4.3) &	45.7 (4.3)	43.9 (4.0)	47.4 (3.8)	45.2 (4.3)
DBP Lead	-43.1 (4.9)	-39.8 (4.5)	-46.9 (4.2)	-43.9 (4.9)	-30.6 (4.9) &	-40.9 (4.5) &	-34.4 (4.2) & @	-31.5 (4.9) & @	-33.7 (4.9) #	-36.2 (4.5) #	-48.2 (4.2)	-51.3 (4.9)
CO	-3.2 (6.5)	4.7 (6.0)	-6.2 (5.6)	-2.3 (6.5)	2.5 (6.5)	2.2 (6.0)	0.4 (5.6)	-2.3 (6.5)	-5.4 (6.5)	-0.3 (6.0)	5.6 (5.6)	7.8 (6.5)
CO Lag	37.0 (4.0) &	44.1 (3.7) &	38.2 (3.5) &	39.7 (4.0) &	32.6 (4.0) &	34.5 (3.7) &	36.4 (3.5) &	32.1 (4.0) &	48.4 (4.0) &	46.1 (3.7) &	51.2 (3.5) &	44.0 (4.0) &

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
CO Lead	-39.4 (5.0)	-28.8 (4.6)	-38.7 (4.3)	-43.1 (5.0)	-32.9 (5.0)	-31.5 (4.6)	-33.0 (4.3)	-34.0 (5.0)	-46.3 (5.0) &	-41.4 (4.6) &	-46.3 (4.3) &	-50.0 (5.0) &
COPx	-39.5 (9.1) & #	2.8 (8.5) & #	-32.7 (7.9) &	-51.2 (9.1) &	-27.2 (9.1) &	-5.5 (8.5) &	-5.1 (7.9) &	-19.4 (9.1) &	-9.7 (9.1) & #	-0.8 (8.5) & #	4.7 (7.9) &	19.9 (9.1) &
COPx Lag	37.5 (5.9) @	52.4 (5.5) @	35.5 (5.1)	32.0 (5.9)	27.7 (5.9) & @	30.5 (5.5) & @	33.7 (5.1) &	25.8 (5.9) &	42.1 (5.9) @	48.9 (5.5) @	34.6 (5.1)	42.0 (5.9)
COPx Lead	-67.5 (5.6) &	-47.0 (5.2) & €	-57.8 (4.8) &	-68.6 (5.6) &	-45.3 (5.6)	-31.4 (5.2) €	-32.7 (4.8)	-43.1 (5.6)	-46.8 (5.6)	-36.3 (5.2) €	-38.2 (4.8)	-25.0 (5.6)
COPy	34.0 (8.0) &	9.6 (7.4) &	28.6 (6.9) &	42.8 (8.0) &	12.1 (8.0)	2.5 (7.4)	7.3 (6.9)	12.1 (8.0)	4.8 (8.0)	-5.7 (7.4)	9.9 (6.9)	4.2 (8.0)
COPy Lag	54.4 (3.9)	47.3 (3.6) **	55.1 (3.3)	60.4 (3.9) €	33.1 (3.9) &	33.0 (3.6) & **	31.3 (3.3) &	33.2 (3.9) & €	61.4 (3.9)	48.9 (3.6) **	42.8 (3.3)	63.7 (3.9) €
COPy Lead	-30.8 (5.2) &	-42.8 (4.8) &	-37.8 (4.5) &	-35.1 (5.2) &	-24.1 (5.2) &	-29.5 (4.8) &	-26.3 (4.5) &	-26.0 (5.2) &	-53.4 (5.2) &	-42.8 (4.8) &	-37.4 (4.5) &	-42.4 (5.2) &

#: significantly different from the values for Old in same frequency;
&: significantly different from the values in the other two frequency bands in all condition;
@: significantly different from the values for old in all conditions;
€ : significantly different from the values for female in same age group;
**: significantly different from the values for old in same gender
Significance level at $p<0.05$;

Table 4-11 Least square mean of average value of coherence > threshold. Comparison between the young and elderly participant group in eyes closed condition in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). (young n = 13; old n = 14). Data reported as Mean (SEM)

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
SBP	0.330 (0.015) #	0.310 (0.014) #	0.339 (0.013) #	0.298 (0.015) #	0.408 (0.015) #	0.402 (0.014) #	0.394 (0.013) #	0.403 (0.015) #	0.449 (0.015) #	0.434 (0.014) #	0.448 (0.013) #	0.447 (0.015) #
DBP	0.352 (0.015) #	0.325 (0.014) #	0.347 (0.013) #	0.322 (0.015) #	0.402 (0.015) #	0.403 (0.014) #	0.401 (0.013) #	0.400 (0.015) #	0.447 (0.015) #	0.438 (0.014) #	0.439 (0.013) #	0.448 (0.015) #
CO	0.343 (0.013) #	0.319 (0.012) #	0.332 (0.011) #	0.293 (0.013) #	0.397 (0.013) #	0.406 (0.012) #	0.403 (0.011) #	0.417 (0.013) #	0.433 (0.013) #	0.461 (0.012) #	0.443 (0.011) #	0.426 (0.013) #
COPx	0.463 (0.027) @	0.466 (0.025)	0.432 (0.023) @	0.506 (0.027)	0.460 (0.027) @	0.506 (0.025)	0.434 (0.023) @	0.470 (0.027)	0.513 (0.027) # @	0.539 (0.025) #	0.489 (0.023) # @	0.561 (0.027) #
COPy	0.436 (0.021)	0.413 (0.020)	0.420 (0.018)	0.490 (0.021) &	0.460 (0.021)	0.443 (0.020)	0.434 (0.018)	0.460 (0.021) &	0.485 (0.021) #	0.472 (0.020) #	0.455 (0.018) #	0.510 (0.021) # &

#: significantly different from values in the other two frequency bands;
@: significantly difference from values for males in all conditions;
&: significantly different from values for Y males and Old females.;
Significance level at $p<0.05$

Table 4-12 Least square mean of the average of variable values in the time period of significant coherence. Comparison between the young and elderly participant group across gender and age in the three frequency bands LF (0.05 - 0.1 Hz); VLF (0.01 – 0.05 Hz); ULF (0.005 – 0.01 Hz). (young n = 13; old n = 14). Data reported as Mean (SEM)

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
EMG	5.36 (0.93) E-03 @	7.29 (0.86) E-03	7.64 (0.80) E-03 @ #	9.20 (0.93) E-03 #	5.39 (0.93) E-03 @	7.34 (0.86) E-03	7.62 (0.80) E-03 @ #	9.31 (0.93) E-03 #	7.77 (0.80) E-03 @ &	1.00 (0.09) E-02 &	5.55 (0.93) E-03 @ # &	7.24 (0.86) E-03 # &
SBP	109 (8)	120 (7)	138 (7) #	146 (8) #	109 (8)	120 (7)	139 (7) #	146 (8) #	139 (7)	147 (8)	109 (8) #	120 (7) #
EMG	5.36 (0.97) E-03	7.28 (0.90) E-03	7.62 (0.84) E-03 #	9.22 (0.97) E-03 #	5.40 (0.97) E-03	7.36 (0.90) E-03	7.57 (0.84) E-03 #	9.33 (0.97) E-03 #	8.01 (0.84) E-03	9.70 (0.97) E-03	5.49 (0.97) E-03 #	7.31 (0.90) E-03 #
DBP	66 (4)	71 (4)	70 (4)	66 (4)	66 (4)	71 (4)	71 (4)	66 (4)	71 (4)	66 (4)	66 (4)	71 (4)
EMG	5.36 (0.93) E-03 @	7.28 (0.86) E-03	7.63 (0.81) E-03 @ #	9.24 (0.93) E-03 #	5.41 (0.93) E-03 @	7.31 (0.86) E-03	7.41 (0.81) E-03 @ #	9.33 (0.93) E-03 #	7.69 (0.81) E-03 @	9.44 (0.93) E-03	5.40 (0.93) E-03 @ #	7.25 (0.86) E-03 #

	LF				VLF				ULF			
	YOUNG		OLD		YOUNG		OLD		YOUNG		OLD	
	F	M	F	M	F	M	F	M	F	M	F	M
CO	4.25 (0.55) @	4.63 (0.51)	6.21 (0.48) @ #	8.72 (0.55) #	4.23 (0.55) @	4.63 (0.51)	6.21 (0.48) @ #	8.79 (0.55) #	6.19 (0.48) @	8.61 (0.55)	4.24 (0.55) @ #	4.66 (0.51) #
EMG	5.36 (0.95) E-03	7.30 (0.87) E-03	7.63 (0.82) E-03 #	9.23 (0.95) E-03 #	5.39 (0.95) E-03	7.32 (0.87) E-03	7.53 (0.82) E-03 #	9.37 (0.95) E-03 #	7.73 (0.82) E-03	9.40 (0.95) E-03	5.39 (0.95) E-03 #	7.33 (0.87) E-03 #
COPx	2.32 (4.37) E-03	-1.20 (0.40) E-02	5.14 (3.78) E-03 #	2.48 (4.37) E-03 #	2.31 (4.37) E-03	-1.20 (0.40) E-02	5.16 (3.78) E-03 #	2.48 (4.37) E-03 #	5.11 (3.78) E-03	2.49 (4.37) E-03	2.28 (4.37) E-03 #	-1.19 (0.40) E-03 #
EMG	5.37 (0.93) E-03	7.28 (0.86) E-03	7.62 (0.80) E-03 #	9.23 (0.93) E-03 #	5.43 (0.93) E-03	7.34 (0.86) E-03	7.50 (0.80) E-03 #	9.29 (0.93) E-03 #	7.96 (0.80) E-03	9.59 (0.93) E-03	5.38 (0.93) E-03 #	7.26 (0.86) E-03 #
COPy	0.41 (35.1) E-03	6.40 (3.25) E-02	0.016 (30.4) E-03	-0.60 (35.1) E-03	0.38 (35.1) E-03	6.40 (3.25) E-02	2.20E-06 (3.04E-02)	-0.59 (35.1) E-03	-0.049 (30.4) E-03 &	-0.68 (35.1) E-03 &	0.34 (35.1) E-03 &	6.36 (3.25) E-02 &

@: significantly different from values for males in all conditions;
#: significantly different from values in the young group in all conditions;
&: significantly different from values in the other two frequency bands.
Significance level at $p<0.05$

4.5 Conclusion.

The cardiovascular and posture control systems were examined for the effects of ageing on the interaction between the two systems. The systolic blood pressure showed an increase in the average value with removal of visual input to the central nervous control. The cardiac output was higher in males than females and showed an increase in value with eyes closed condition. The phase lead of the cardiovascular variables was significantly higher in the ultra low frequency than the other two bands indicating a possibility of longer time delay in the cause and effect relation.

The switch time from short term to long-term strategy for posture control was significantly lower in the older compared to the younger age group. This would indicate that the elderly individuals have a faster switch to, and reliance on, long-term stability than immediate short-term stability. The short-term scaling exponent increased and long-term exponent decreased in value with eyes closed. This implies for the higher tendency towards short-term stability with removal of visual input. Hence, the closed loop control is reduced on the short term basis and is increased on the long term basis.

The systolic blood pressure and cardiac output was higher in elderly individuals in comparison to young individuals, possibly attributed to the change in vasculature and central components of the systems. The average value for posture muscle activity (EMG) was significantly higher for elderly individuals than young individuals, which suggests increased muscle activation for the maintenance of posture and orthostasis.

In, summary the analysis has shown that there is a significant change in the interaction dynamics between the posture and cardiovascular controls with age. The elderly age group showed increased coupling between the two systems and higher values of characteristic variables suggesting that the physiological changes occurring with age are associated with higher systemic activation levels.

CHAPTER 5: GENERAL CONCLUSION

5.1 Overview

Falls are a leading cause of injury and deaths in the elderly. It is amongst the largest health expenditures in Canada. A large number of these falls occur due orthostatic intolerance leading to postural fainting. The aim of the three studies conducted in this thesis was to establish the existence of a relationship between the cardiovascular and postural control systems and to characterize the effects of ageing on the relationship in healthy idividuals. The interrelationship in the healthy section of elderly age group was established to provide a baseline to understand the behaviour in the non-fallers group of elderly individuals.

The first study examined the wavelet transform coherence (WTC) method for its applicability and effectiveness in analysis of non-stationary signal sets. The method showed low bias and error rates, which provided confidence in its applicability to the signal sets under evaluation. The coherence thresholds and baseline statistics of the coherence estimate as a time- frequency map established WTC as a promising candidate for non-stationary physiological signal analysis.

The second study analyzed data collected from healthy young individuals for existence of a relationship between the cardiovascular and postural systems. It was hypothesized that a bidirectional relationship existed between the cardiovascular and postural controls. The WTC method detected periods of

significant coherence and phase locking between the signals over the duration of four minutes of quiet stance. These data strongly suggests a possible interaction between the cardiovascular and postural control systems and provides further opportunity for investigation.

The third study investigated the effects of ageing on the bi-directional relationship observed in the second study. The interrelationship in the healthy non-fallers group in the elderly was studied to establish the ideal characteristics in the elderly age group. The interaction dynamics (cardiovascular-postural-cardiovascular) in the older age group were similar to the young age group; however, there was increased magnitude of the cardiovascular variables in the elderly group. The presence of no significant change of the transfer characteristics with age and the observation of increased activation of the posture muscles (EMG) in the lower extremity in the elderly; indicates that these muscle groups are used in parallel for the maintenance of upright posture and BP through the skeletal muscle pump. The increased level of EMG compensates for the physiological changes in the cardiovascular system due to age.

In summary, the present thesis has successfully exhibited the existence of bi-directional relationship between the postural and cardiovascular controls. With an increase in age, the interaction presented with similar behaviour of the young age group; however, the individual systems had increased activity. The thesis has successfully characterized the effect of ageing on the cardio-postural interaction by studying the healthy segment of the elderly age group.

5.2 Difficulties faced

The understanding of the data acquisition methods for EMG (signal acquisition, electrode placement) and stabilometry (center of pressure trajectory) was a new endeavour and involved thorough literature review and practical testing taking a substantial time for completion of the pilot work.

The mathematical understanding and statistical validation of the wavelet transform based coherence estimator took about 3 months time involving a continuous computation time of 3 weeks.

Handling of data from 30 participants was a cumbersome task and lead to erroneous calculations and misinterpretation. These had to be corrected and lead to more time investment into the thesis work.

5.3 Limitations

Other time frequency analysis methods, such as Hilbert transform and variants of the wavelet method were not evaluated for their applicability due to time constraint. These methods, often used in neuroscience could present new paradigms when applied to cardiovascular and postural systems research.

The exclusion criteria for the recruitment of elderly participants were stringent though theoretically valid, and resulted in a small participant pool. The limited size of the participant pool affected the power of the statistical tests.

The usable data for comparison was even less due to the equipment, data acquisition, and experimenter errors.

Environmental conditions such as room temperature and ambient noise were not controlled and may have resulted in additional sensory cues and could have interfered with the central nervous system control.

5.4 Future work

The future work should follow from the results of the present study and investigate the exact dependence between the systems. The task would involve establishing the control system model for the interaction of the cardiovascular and posture control system and central nervous control centres. Each sensory input and the corresponding control centre needs to be studied separately in order to understand the contribution and the overall effect of the contribution on the holistic effect on the system.

The present work analyzed the interdependence in the three frequency bands, being an initial analysis. Future work can investigate the existence the interrelationship and its behaviour in the same frequency region but better bandwidth definition. This can further add to the analysis of the integrated control system mentioned above.

Proven statistical methods such as principal component analysis and independent component analysis can be utilized to identify the components having a one to one interaction, rather than a holistic approach.

In order to take the physiological knowledge to real life situations, the diseased models need to be understood and investigated for the effect on the

established bidirectional relationship between the cardiovascular and postural control.

The present thesis investigated the effects of ageing in a healthy segment of elderly individuals. Further study of the interrelationship between the two systems in the fallers group of elderly will provide us indications for the state of the systems and the interrelationship that induces the falls.

Clinically relevant parameters need to derived from the models and interaction dynamics which can translate into diagnostic tools, prognosis methodologies and treatment protocols

CHAPTER 6: APPENDICES

APPENDIX A

The work presented here was done as a pilot study before proceeding into the main thesis idea. The method of Wavelet transform coherence and stabilogram diffusion analysis stemmed out after certain shortcomings in the analysis presented below.

Relationship between the EMG, COP and BP signals

Time series analysis was applied to find the relationship between the three signals. The total EMG activity of the legs was calculated by adding the rectified EMG activity from the gastrocnemius muscles of both the legs. The envelop EMG and BP signals were obtained by low pass filtering the signals at 1Hz cut off frequency.

Correlation analysis was performed between EMG to BP and COP to BP for segments of 15 second duration traversing the whole data set at a step of 1 second (Figure 6-1). Last 4 minutes duration in quiet stance with eyes closed were analyzed. The correlation analysis showed many regions with significant correlation ($r>0.5$, $p<0.0001$) indicating the presence of an interdependence between EMG to BP and COP to BP. As an example, signals in segments 46 and 164 are plotted to show the positive and negative relation between them (Figure 6-2, 6-3).

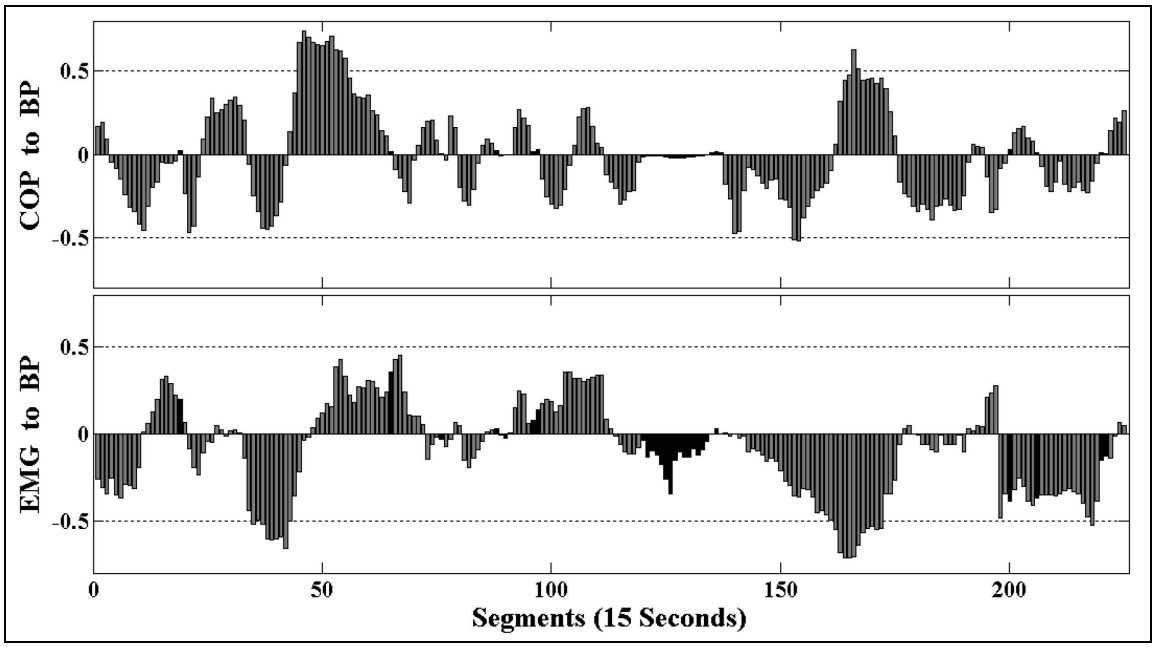

Figure 6-1 Correlation between COP and BP, and EMG and BP for segments of 15 seconds length with a step of 1 second (Total = 225). Grey color represents significant correlation (p<0.0001) and black represents non significant correlation. Segment zero is the start of the last 4 minutes in the stance phase with eyes closed

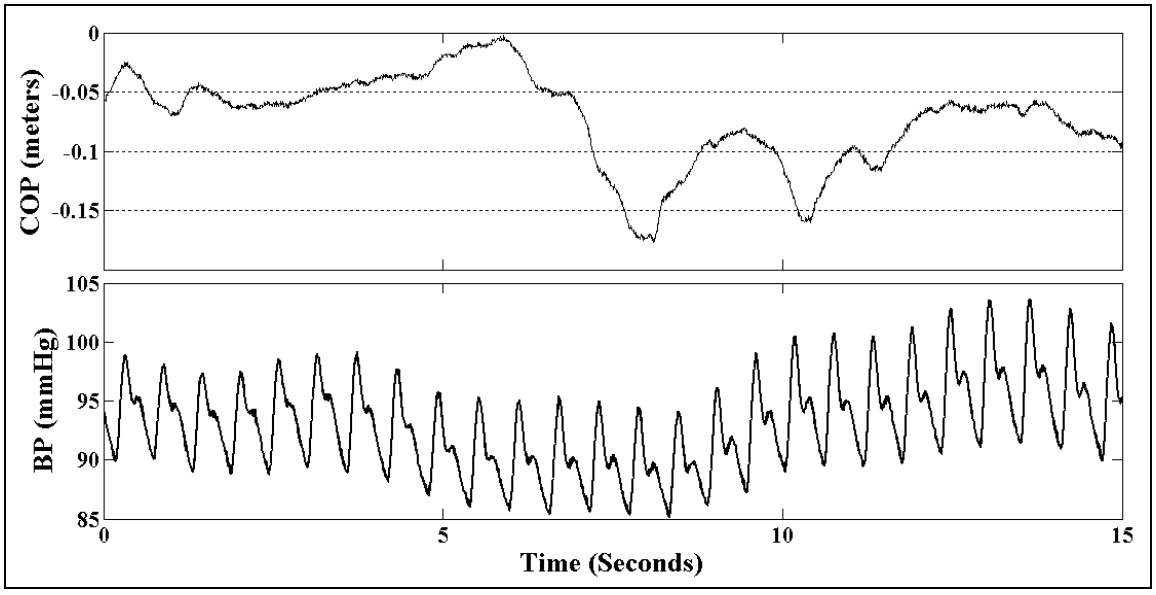

Figure 6-2 The COP and BP signals for segment number 46 show a positive dependence on each other (r>0.5, p<0.0001).

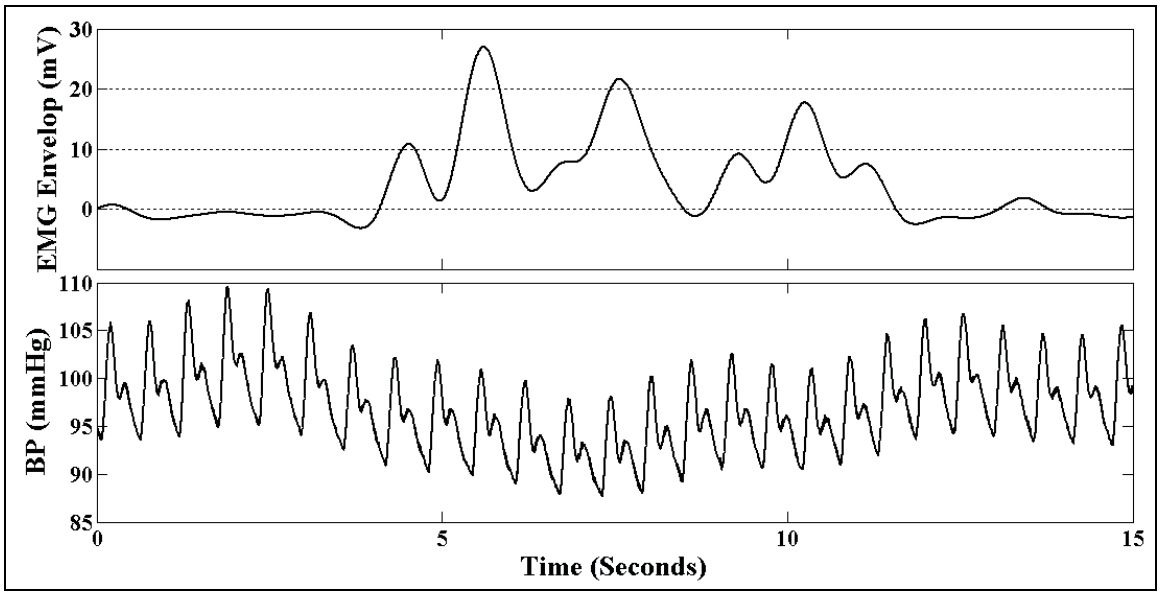

Figure 6-3 The EMG and BP signals for segment number 164 show a negative dependence on each other (r>0.5, p<0.0001).

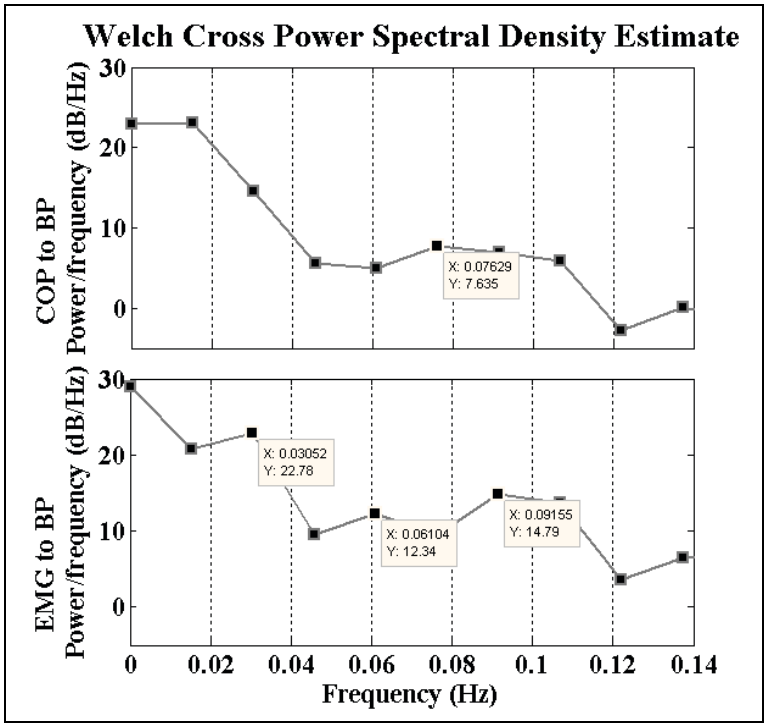

Figure 6-4 The CPSD estimate between COP and BP, and EMG and BP indicates peaks in the low frequency range of 0.03 Hz to 0.09 Hz

Figure 6-4 above shows Cross Power Spectral Density plots for the signals in last 4 minutes of quiet stance with eyes closed. Higher number of

peaks between EMG to BP in the frequency range 0-0.1 Hz indicate closer coupling between them in comparison with the COP to BP coupling.

Wavelet approach: relationship between the EMG, COP and BP signals

Wavelet decomposition was applied using Discrete Wavelet Transform with Daubechies5 wavelet. Approximation signal in the frequency range of ~0.11Hz was obtained by decomposition of the signal resampled at 15 Hz to the 8th level.

Individual leg EMGs were considered separately and analysis was based on the blood pressure variation to EMG activity. The correlation analysis applied to the wavelet reconstructed signals showed regions of significant high correlation ($r>0.5$, $p<0.0001$) (Figure 6-5). The correlation plots also showed alternating selective excitation of each leg contributing to the blood pressure change. There were regions where each leg showed activity of opposite nature indicating contraction and relaxation in either of them at the same time. In some regions it also showed behaviour of same nature indicating co-contraction of the two legs.

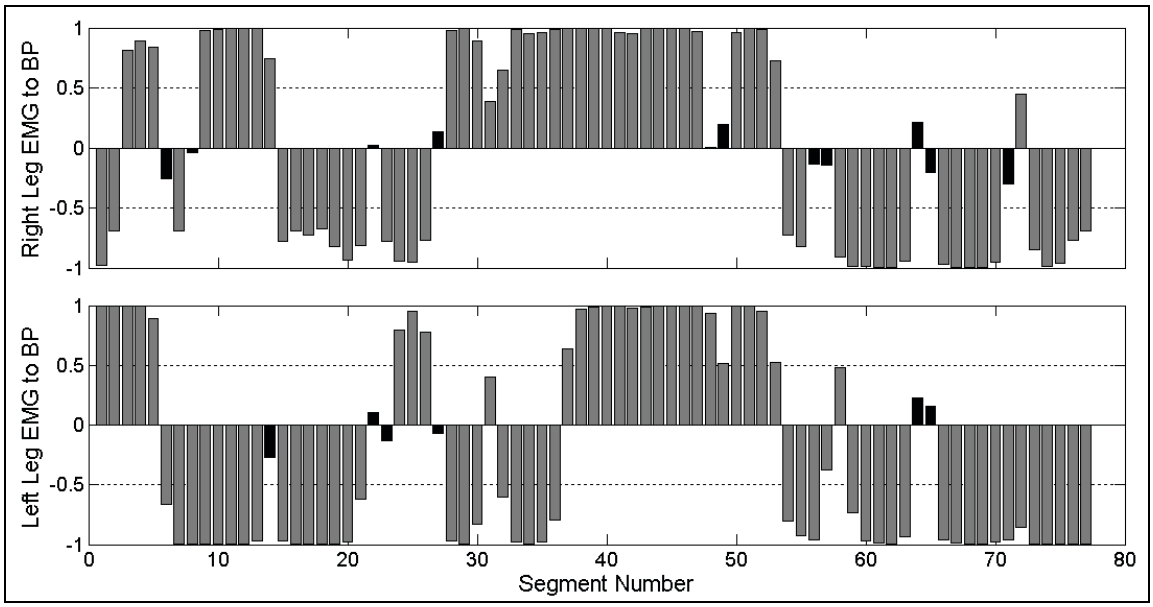

Figure 6-5 Correlation analysis for the wavelet reconstructed individual leg EMG to BP signals. Grey colour represents significant correlation (p<0.0001) and black represents non-significant correlation.

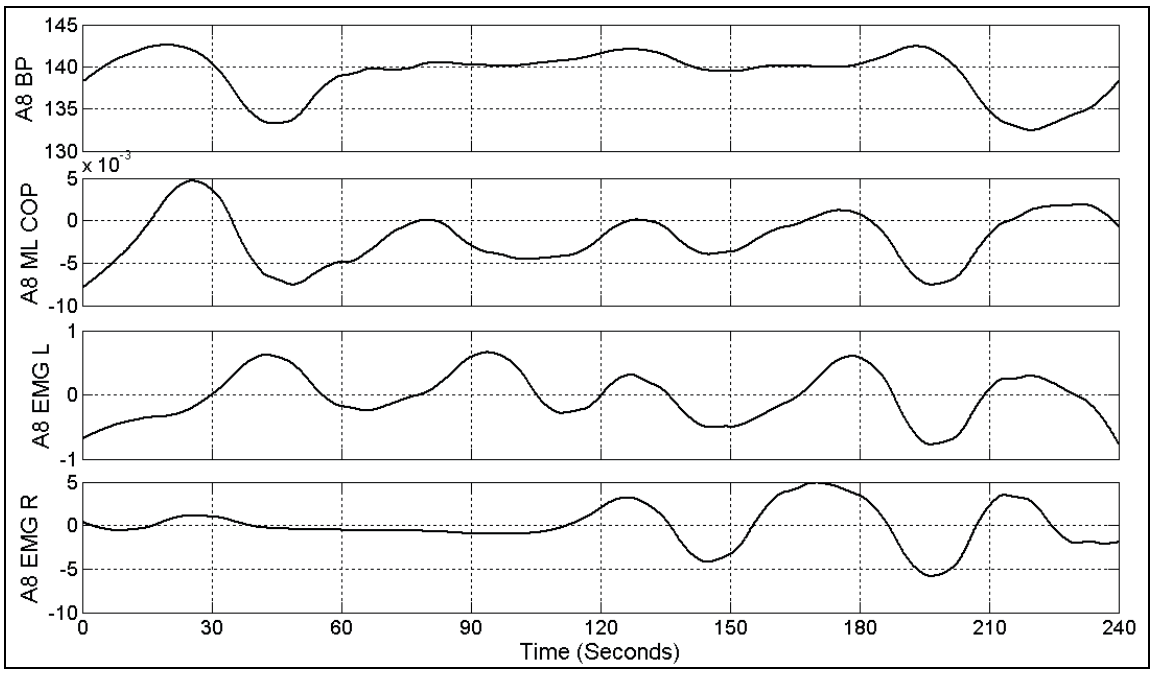

Figure 6-6 Wavelet reconstructed approximation signals (8th Level) for COP, BP and individual leg EMG.

Figure 6-6 above shows the 8th scale wavelet reconstructed signals for the last 4 minutes duration of quiet stance with eyes closed condition.

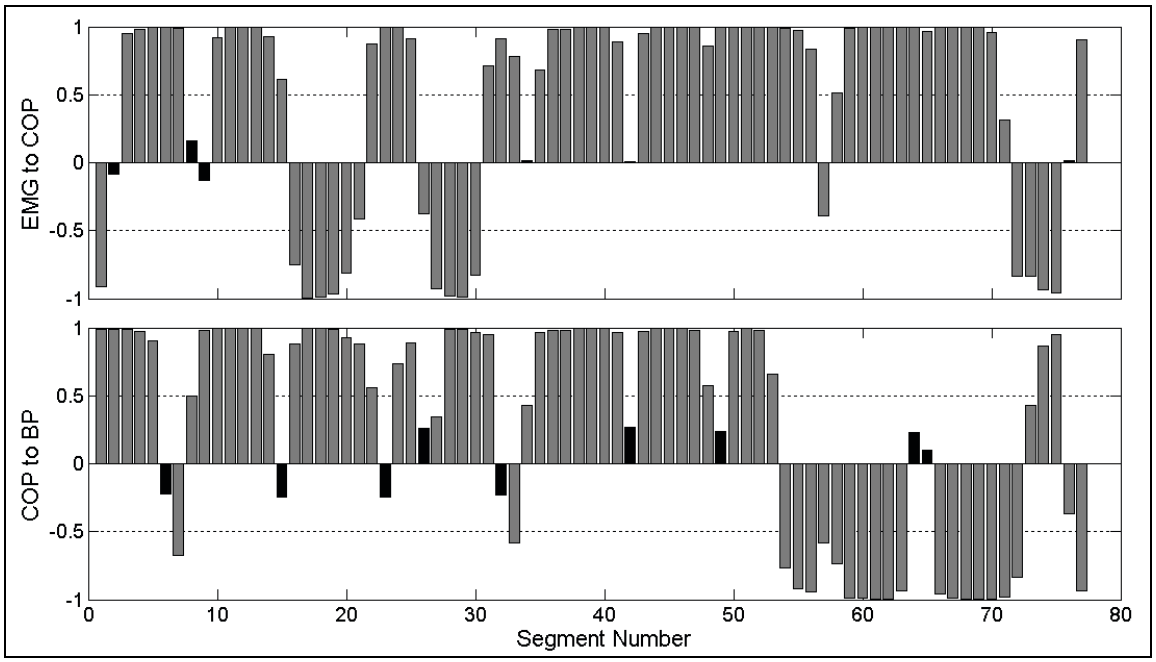

Figure 6-7 Correlation analysis of wavelet reconstructed COP to BP and total EMG to COP. Grey color represents significant correlation ($p<0.0001$) and black represents non significant correlation.

APPENDIX B

Publications during the M.Sc. Program (Sept 2008- Dec 2010)

Amanmeet Garg, Philippe A. Souvestre, Stephen Robinovitch, Andrew P. Blaber, “Relationship Between Gastrocnemius Electromyographic Activity and Blood Pressure Variation with Respect to Postural Sway During Quiet Stance”, 7th Edition of Progress in Motor Controls, Marseille, France. July 22-25, 2009

Amanmeet Garg, Andrew P. Blaber; “An investigation of blood pressure change with posture muscle activation: a wavelet approach”; Federation of European Physiological Societies meeting 2009, November 12-16, Lubljana, Slovenia.

Philippe A. Souvestre, **Amanmeet Garg**, Andrew P. Blaber, “Can loss of spatial reference link concussion to both anxiety and gait disorders? Introducing a predictive integrative assessment model”; 3rd International Congress on Gait & Mental Function, Washington, USA, February 26-28, 2010.

Brett Hollowell, Michelle Bruner, **Amanmeet Garg**, Andrew P. Blaber; “Efficacy of combined HVR and HCVR tests for assessment of oxygen and carbon dioxide sensitivity.”; Aerospace Medical Association 81st Annual Scientific Meeting, May 9-13, 2010, Phoenix AZ, USA.

Amanmeet Garg, Michelle Bruner, Philippe A. Souvestre, Andrew P. Blaber; “Plantar biophotonic stimulation improves ocular-motor and postural control in motor vehicle accident patients”; The 33rd conference of the Canadian medical and biological engineering society, June 15-18, 2010, Vancouver, BC, Canada.

Amanmeet Garg, Andrew P. Blaber, “Investigation of the relationship between blood pressure and postural electromyography signals using wavelet transform coherence methodology”, In preparation.

REFERENCE LIST

Addison, P. S. (2005) Wavelet transforms and the ECG: a review. Physiol. Measure. 26, R155–R199.

Aniansson A, Hedberg M, Henning GB, Grimby G (1986) Muscle morphology, enzymatic activity, and muscle strength in elderly men: a follow-up study. Muscle Nerve 9:585–591

Baily RG, Prophet SA, Shenberger JS, Zelis R and Sinoway LI.(1990) Direct neurohumoral evidence for isolated sympathetic nervous system activation to skeletal muscle in response to cardiopulmonary baroreceptor unloading. Circ Res 66:1720-1728.

Bendat, S. J., Piersol, G. A. (1971) Random data: analysis and measurement procedures. New York, NY: Wiley.

Blaber AP, Yamamoto Y, Hughson RL. (1995) Methodology of spontaneous baroreflex relationship assessed by surrogate data analysis. American Journal of Physiology;268:H1682–7.

Blaber, A.P., Hartley, T., Pretorius, P.J. (2003) Effect of acute exposure to 3,660 m altitude on orthostatic responses and tolerance. J. Appl. Physiol. 95:591-601.

Blaber A.P., Landrock C.K., Souvestre P.A.,(2009) Cardio-postural deconditioning: A model for post-flight orthostatic intolerance, Respiratory Physiology & Neurobiology, Volume 169, Supplement 1, Cardio-Respiratory Physiology in Space, Pages S21-S25, ISSN 1569-9048.

Black F.O., Nashner L.M.(1985), Postural control in four classes of vestibular abnormalities. In: Igarashi M., Black F.O.: Vestibular and visual control on posture and locomotor equilibrium. Karger, Basel, 271

Blaszczyk JW, Lowe DL, Hansen PD. (1994), Ranges of postural stability and their changes in the elderly. Gait Posture; 2: 11–7.

Bonato, P.; Roy, S.H.; Knaflitz, M.; De Luca, C.J.; (2001) Time-frequency parameters of the surface myoelectric signal for assessing muscle fatigue during cyclic dynamic contractions, IEEE Transactions on Biomedical Engineering, vol.48, no.7, pp.745-753.

Brauer SG, Burns YR, Galley P.(2000) A prospective study of laboratory and clinical measures of postural stability to predict community-dwelling fallers. J Gerontol A Biol Sci Med Sci. 55(8):M469-76.

Bruce, M. (1980). The relation of tactile thresholds to histology in the fingers of elderly people. Journal of Neurology, Neurosurgery and Psychiatry, 43, 730-734.

Buatois S, Gueguen R, Gauchard GC, Benetos A, Perrin PP (2006). Posturography and risk of recurrent falls in healthy non-institutionalized persons aged over 65. Gerontology. 52(6):345-52.

Chiari L, Cappello A, Lenzi D, Croce UD, (2000) An improved technique for the extraction of stochastic parameters for stabilograms, Gait and Posture 12 225-234.

CIHI (2004). National Trauma registry 2004 report: injury hospitalizations. Ottawa Canadian Institute of health information

Claydon VE, Hainsworth R,(2005); Increased postural sway in control subjects with poor orthostatic tolerance, J. Am. Coll. Cardiol.; 46;1309-1313;

Dichgans J, Diener H.C. (1989), The contribution of vestibulo-spinal mechanisms to the maintenance of human upright posture, Acta Otolaryngol. 107 (5–6) 338–345.

Farge, M. (1992) Wavelet transforms and their applications to turbulence. Annual Reviews of Fluid Mechanics. 24, 395–457.

Fisher L. D., Van Belle G., (1993) Biostatistics. A Methodology for Health Sciences. New York: Wiley.

Gagey P.M. (1991), Non-Vestibular dizziness and static posturography, Acta oto-rhino-laryngologica belg., 45, 335-339

Grinsted, A., Moore, J. C., Jevrejeva, S. (2004) Application of the cross wavelet transform and the wavelet coherence to geophysical time series. Nonlinear Processes in Geophysics. 11, 561–566.

Hachiya T, Blaber AP, Saito M. (2008), Near infrared spectroscopy provides an index of blood flow and vasoconstriction in calf skeletal muscle during lower body negative pressure. Acta Physiol, 193, 117–127

Hermens HJ, Freriks B, Merletti R, Hagg G, Stegeman D, Blok J et al editors.(1999) SENIAM 8: European recommendations for surface electromyography, ISBN: 90-75452-15-2: Roessing Research and Development.

Horak FB, Macpherson JM. (1996) Postural orientation and equilibrium. In: Shepard J & Rowell L, eds.Handbook of physiology, section 12. Exercise: regulation and integration of multiple systems. New York, Oxford University, 1996: 255-292

Hsiao-Wecksler, E.T., Robinovitch, S.N.(2007): The effect of step length on young and elderly women's ability to recover balance. Clinical Biomechanics, 22: 574–80.

Hurely MV, Rees J, Newham DJ. (1998) Quadriceps function, proprioceptive acuity and functional performance in healthy young, middle-aged and elderly subjects. Age Ageing 1998; 27: 55–62.

Jacobsen TN, Morgan BJ, Scherrer U, Vissing SF, Lange RA, Johnson N, Ring WS, Rahko PS, Hanson, P, Victor RG.(1993) Relative contributions of cardiopulmonary and sinoaortic baroreflexes in causing sympathetic activation in the human skeletal muscle circulation during orthostatic stress. Circ Res 73(2): 367-378.

Janssen I, Heymsfield SB, Wang Z, Ross R (2000); Skeletal muscle mass and distribution in 468 men and women aged 18-88 year old; J Appl Physiol; 89:81-88

Jardine DL, Ikram H, Frampton CM, Frethey R, Bennett SI, Crozier IG (1998). Autonomic control of vasovagal syncope. Am J Physiol 274: H2110-H2115.

Joseph J., Nightingale A., Williams P.L. (1955), A detailed study of electric potentials recorded over some postural muscles while relaxed and standing. J. Physiol I27, 6I7-625, 1955.

Kay, S. M. (1988) Modern spectral estimation: theory and application. Upper Saddle River, NJ:Prentice Hall.

Keissar K, Davrath LR, Akselrod S. (2009) Coherence analysis betweenrespiration and heart rate variability using continuous wavelet transform. Phil Trans R Soc A; 367:1393–406

Kilby J., Hosseini H. G., (2004) Wavelet analysis of surface electromyography signals, Proceedings of the 26th annual international conference of the IEEE EMBS, pp. 384–387.

Kilby, J.; Mawston, G.; Hosseini, H. G.; (2006) Analysis of Surface Electromyography Signals using Continuous Wavelet Transform for Feature Extraction, Advances in Medical, Signal and Information Processing, 2006. MEDSIP 2006. IET 3rd International Conference On , vol., no., pp.1-4, 17-19

Kimmerly DS, Shoemaker JK.(2002) Hypovolemia and neurovascular control during orthostatic stress. Am. J Physiol Heart Circ Physiol 282(2): H645-H655.

Klein, A., Sauer, T., Jedynak, A., Skrandies, W. (2006) Conventional and wavelet coherence applied to sensory-evoked electrical brain activity. IEEE Transactions on Biomedical Engineering. 53, 266–272.

Kumar D. K., Pah N. D., Bradley A., (2003) Wavelet analysis of surface electromyography to determine muscle fatigue, IEEE Transactions on neural systems and rehabilitation engineering, vol. 11, pp. 400–406.

Lachaux, J.-P., Lutz, A., Rudrauf, D., Cosmelli, D., Le Van Quyen, M., Martinerie, J., Varela, F. (2002). Estimating the timecourse of coherence between single-trial brain signals: an introduction to wavelet coherence. Neurophysiologie Clinique, 32, 157-174

Laszlo Z, Rossler A, Hinghofer-Szalkay HG(1998), Cardiovascular changes during and after different LBNP levels in Men. Aviat Space Environ Med 69: 32-39.

Lewis T. A (1932), lecture on vasovagal syncope and the carotid sinus mechanism: with comments on Gower's and Nothnagel's syndrome. Br Med J 1: 873- 876,.

Lin SI, Woollacott MH, Jensen JL. (2004), Postural responses in older adults of different levels of functional balance ability. Aging Clin Exp Res; 16: 369–74.

Loram I.D., Maganaris C.N., Lakie M.; (2005) Human postural sway results from frequent, ballistic bias impulses by soleus and gastrocnemius Journal of Physiology 564 (1) 295-311.

Lord SR, Clark RD, Webster IW.(1991) Postural stability and associated physiological factors in a population of aged persons. Gerontol; 46: M69–76.

Lundvall J, Bjerkhoel P, Edfeldt H,Ivarsson C and Lanne T.(1993) Dynamics of transcapillary fluid transfer and plasma volume during lower body negative pressure. Acta Physiol Scand 147: 163-172.

Mackey, D.C., Robinovitch, S.N.(2006): Mechanisms underlying age-related differences in ability to recover balance with the ankle strategy, Gait and Posture, 23(1): 59-68.

Maraun D Kurths J (2004) Cross wavelet analysis: significance testing and pitfalls. Nonlinear Proc Geoph. 11: 505-514

Marrone A, Polosa A D, Scioscia G, Stramaglia S, Zenzola A. (1999) Wavelet analysis of blod pressure waves in vasovagal syncope. Physica A 271:458-469.

McCollum G, Shupert CL, Nashner LM (1996) Organizing sensory information for postural control in altered sensory environments. J Theor Biol 180:257–270

McKeown M. J., Radtke R. (2001). Phasic and tonic coupling between EEG and EMG demonstrated with independent component analysis. Journal of Clinical Neurophysiology, 18, 45–57.

Melzer I, Benjuya N, Kaplanski J.(2004) Postural stability in the elderly: a comparison between fallers and non-fallers. Age Ageing. 33(6):602-7.

Mitro P., Hijova E.(2006) Myocardial contractility and cardiac filling measured by impedance cardiography in patients with nitroglycerine-induced vasovagal syncope. Pacing Clin Electrophysiol 29: 1-8.

Mohanty PK, Thames MD, Arrowood JA, Sowers JR, McNamara C, and Szentpetery S.(1987) Impairment of cardiopulmonary baroreflex after cardiac transplantation in humans. Circ 75: 914-921.

Nevitt MC, Cummings SR, Kidd S, Black D.(1989) Risk factors for recurrent nonsyncopal falls: a prospective study. JAMA; 261:2663-8.

Nevitt MC, Cummings SR, Hudes ES(1991), Risk factors for injurious falls: a prospective study. J Gerontol;46:M164-M170

Nowak L, Nowak FG, Janko S, Dorwarth U, Hoffmann E and Botzenhardt F. (2007), Investigation of various types of neurocardiogenic response to head-up tilting by extended hemodynamic and neurohumoral monitoring. Pacing Clin Electrophysiol 30: 623-630.

O'Leary DD, Kimmerly DS, Cechetto AD and Shoemaker JK.(2003), Differential effect of head-up tilt on cardiovagal and sympathetic baroreflex sensitivity in humans. Exp Physiol 88: 769-774.

Padmanabhan, P.; Puthusserypady, S.; (2004) Nonlinear analysis of EMG signals - a chaotic approach, Engineering in Medicine and Biology Society, 2004. IEMBS '04. 26th Annual International Conference of the IEEE , vol.1, no., pp.608-611, 1-5 Sept. 2004.

Paillard, Jacques (1976), Tonus, posture et mouvements. C. Krayser (Editor) Physiologies, Tome 2, Flammarion, Paris, 521-728.

Patla, A.E., Winter D.A., Frank J.S., Walt S.E., & Prased S. (1990). Identification of age-related changes in the balance control system. In: Duncan P, ed. Balance:Proceedings of the APTA Forum., Alexandria, VA:APTA, 43-55

Pawelczyk JA and Raven PB. (1989), Reductions in central venous pressure improve carotid baroreflex responses in conscious men. Am J Physiol 257 (Heart Circ Physiol 26): H1389-H1395.

Pinna, G. D., Maestri, R. (2001) Reliability of transfer function estimates in cardiovascular variability analysis. Med. Biol. Eng. Comput. 39, 338–347.

Porta, A., Furlan, R., Rimoldi, O., Pagani, M. & van de Borne, P. (2002) Quantifying the strength of the linear causal coupling in closed loop interacting cardiovascular variability signals. Biol. Cybern. 86, 241–251.

Robinson T and Potter J.(1997), Cardiopulmonary and arterial baroreflex-mediated control of forearm vasomotor tone is impaired after acute stroke. Stroke. 28: 2357-2362.

Rosenthall U, Rubin W (1975), Degenerative changes in the human vestibular sensory epithelia. Acta Otolaryngol (Stockh) 79:67-80.

Rothwell, J.C. (1994). Control of human voluntary movement. London, Chapman Hall.

Rowell, L.B. (1993) Human Cardiovascular Control, Oxford University Press, Oxford.

Saul, J. P., Berger, R. D., Albrecht, P., Stein, S. P., Chen, M. H. & Cohen, R. J. (1991) Transfer function analysis of the circulation: unique insights into cardiovascular regulation. Am. J. Physiol. Heart Circ. Physiol. 261, H1231–H1245.

Sheldon J H.(1963) The effect of age in the control of sway. Gerontol Clin; 5: 129-138

SMARTRISK (1998). The economic burden of unintentional injury in Canda. Toronto

Stefanovska A, Bracic M, Kvernmo HD. (1999) Wavelet analysis of oscillations in the peripheral blood circulation measured by laser Doppler technique. IEEE Transactions on Biomedical Engineering ;46(10):1230–39.

Stulen, Foster B.; De Luca, Carlo J.; (1981) Frequency Parameters of the Myoelectric Signal as a Measure of Muscle Conduction Velocity, IEEE Transactions on Biomedical Engineering, vol.BME-28, no.7, pp.515-523.

Theiler J., Eubank S., Longtin A., Galdrikian B., Farmer J.D., (1992) Testing for nonlinearity in time series – the method of surrogate data, Physica D, vol 58, pp. 77-94.

Tinetti ME, Speechley M, Ginter SF.(1988), Risk factors for falls among elderly persons living in the community. N Engl J Med; 319: 1701–07.

Tinetti ME.(1994), Prevention of falls and fall injuries in elderly persons: a research agenda. Prevent Med.;23:756–762.

Tinetti ME, Baker D, McAvay G.(1994), A multifactorial intervention to reduce the risk of falling among elderly people living in the community. N Engl J Med; 331: 821–7.

Tinetti ME, Doucette J, Claus E, Marottol RA. (1995), Risk factors for serious injury during falls by older persons in the community. J Am Geriatr Soc;43:1214-21.

Tinetti ME. (2003), Preventing falls in elderly persons. N Engl J Med; 348: 42–49.

Torrence C., Compo G. P. (1998) A practical guide to wavelet analysis. Bulletins of American Meteorological Society 79, 61–78.

Torrence C., Webster P. (1999) Interdecadal changes in the ESNO-monson system. Journal of Climate 12, 2679–2690.

Toupet, M., Gagey, P. M., & Heuschen, S. (1992). Vestibular patients and aging subjects lose use of visual input and expend more energy in static postural control. In: B. Vellas, M. Toupet, L. Rubenstein, J. L. Albarede & Y. Christen (Eds.), Falls, balance and gait disorders in the elderly (pp 183-198). Paris: Elsevier

Tsai, M.L., Shann, W.C., Luo, W.R., Yen, C.T., (2004). Wavelet-based analysis of low-frequency fluctuations of blood pressure and sympathetic nerve activity in rats. Neurosci. Lett. 358 (3), 165– 168.

Van Milligen, B. Ph., Hidalgo, H., Sanchez, E., Pedrosa, R., Balbin, R., Garcia-Cortes, I. & Tynan, G. R. (1997) Statistically robust linear and nonlinear wavelet analysis applied to plasma edge turbulence. Rev. Sci. Instrum. 68, 967–970.

Voss, A., Schulz, S., Schroeder, R., Baumert, M. & Caminal, P. (2009) Methods derived from nonlinear dynamics for analysing heart rate variability. Phil. Trans. R. Soc. A 367, 277–296.

Wilkins K, (1999), Health care consequences of falls for seniors. Health Report (Statistics Canada, Catalogue 82-03); 10(4):47-55

Winter DA, Prince F, Frank JS, Powell C, Zabjek KF. (1996), Unified theory regarding A:P and M:L balance in quiet stance. J Neurophysiol; 75:2334–43

Woollacott M, Shumway-Cook A (1990) Changes in posture control across the life span: A systems approach. Physical Therapy 70: 799–807