A Wearable Mechatronic Device for Hand Tremor Monitoring and Suppression: Development and Evaluation

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Graduate Program in Biomedical Engineering
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A Wearable Mechatronic Device for Hand Tremor Monitoring and Suppression: Development and Evaluation

Yue Zhou
Ph. D. Thesis, 2019
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Abstract

Tremor, one of the most disabling symptoms of Parkinson’s disease (PD), significantly affects the quality of life of the individuals who suffer from it. These people live with difficulties with fine motor tasks, such as eating and writing, and suffer from social embarrassment. Traditional medicines are often ineffective, and surgery is highly invasive and risky. The emergence of wearable technology facilitates an externally worn mechatronic tremor suppression device as a potential alternative approach for tremor management. However, no device has been developed for the suppression of finger tremor that has been validated on a human.

It has been reported in the literature that tremor can be selectively suppressed by mechanical loading. Therefore, the objectives of this thesis were to develop a wearable tremor suppression device that can suppress tremor at the wrist and the fingers, and to evaluate it on individuals with PD in a pre-clinical trial. To address these objectives, several experiments were performed to quantify hand tremor; an enhanced high-order tremor estimator was developed and evaluated for tremor estimation; and a wearable tremor suppression glove (WTSG) was developed to suppress tremor in the index finger metacarpophalangeal (MCP) joint, the thumb MCP joint, and the wrist.

A total of 18 individuals with PD were recruited for characterizing tremor. The frequencies and magnitudes of the linear acceleration, angular velocity, and angular displacement of tremor in the index finger MCP joint, the thumb MCP joint, and the wrist were quantified. The results showed that parkinsonian tremor consists of multiple harmonics, and that the second and third harmonics cannot be ignored. With the knowledge of the tremor characteristics, an enhanced high-order tremor estimator was developed to acquire better tremor estimation accuracy than its lower-order counterpart.
In addition, the evaluation of the WTSG was conducted on both a physical tremor simulator and on one individual with PD. The results of the simulation study proved the feasibility of using the WTSG to suppress tremor; and the results of the evaluation on a human subject showed that the WTSG can suppress tremor motion while allowing the user to perform voluntary motions. The WTSG developed as a result of this work has demonstrated the feasibility of managing hand tremor with a mechatronic device, and its validation on a human subject has provided useful insights from the user’s perspectives, which facilitate the transition of the WTSG from the lab to the clinic, and eventually to commercial use.

Lastly, an evaluation studying the impact of suppressed tremor on unrestricted joints was conducted on 14 individuals with PD. The results showed a significant increase in tremor magnitude in the unrestricted distal joints when the motions of the proximal joints were restricted. The average increase of the tremor magnitude of the index finger MCP joint, the thumb MCP joint, the wrist and the elbow are 54%, 96%, 124%, and 98% for resting tremor, and 50%, 102%, 49%, and 107% for postural tremor, respectively. Such a result provided additional clinical justification for the significance of the development of a wearable mechatronic device for hand tremor management. Although the focus of this thesis is on hand tremor management, the development and evaluation of a full upper-limb tremor suppression device is required as a future step, in order to advance the use of wearable mechatronic devices as one of the valid tremor treatment approaches.

**Index terms** — Wearable tremor suppression glove, Parkinson’s disease, pathological tremor, hand tremor, tremor suppression, assistive device, tremor characterization, tremor estimation
Summary for Lay Audience

Tremor, one of the most disabling symptoms of Parkinson’s disease (PD), significantly reduces the quality of life of the individuals who are living with tremor, creating difficulties for fine motor tasks, such as eating and writing, and causing social embarrassment. Traditional medicines are often ineffective, and surgery is highly invasive and risky. In addition to these two approaches, an externally worn tremor suppression device has become a potential alternative approach for tremor management. However, no device has been developed for the suppression of finger tremor and validated it on humans.

It has been validated in the literature that tremor can be selectively suppressed by applying force. The objectives of this project were to develop a device that can suppress tremor at the wrist and the fingers, and to evaluate it on individuals with PD. To address these objectives, several experiments were performed to quantify the tremor in the hand; a tremor estimation technique was developed and evaluated; and a wearable tremor suppression glove (WTSG) was developed to suppress tremor in the index finger knuckle, the thumb knuckle and the wrist.

A total of 18 patients were recruited in order to characterize tremor. The frequencies and strength of the tremor in the index finger knuckle, the thumb knuckle and the wrist were quantified. With the knowledge of the tremor characteristics, a high-order tremor estimator showed an average of 28.5% increase in tremor estimation accuracy over its lower-order version.

Lastly, the evaluation of the WTSG was conducted on both a physical tremor simulator and on one individual with PD. The results of the simulation study proved the feasibility of using the WTSG to suppress tremor; and the results of the evaluation on a human subject showed that the WTSG can suppress tremor motion while allowing the voluntary motion of the user. Results from the experiment showed an overall suppression of 73.1%, 80.7% and 85.5% in resting tremor, 70.2%, 79.5% and 81% in postural tremor, and 52.6%, 55.5% and 59.8% in kinetic tremor in the index finger MCP joint, the thumb MCP joint and the wrist, respectively.
To my dearest parents:

Zhenkuan Zhou and Jinming Li
Statement of Co-Authorship

The work presented herein has been written by Yue Zhou under the supervision of Dr. Ana Luisa Trejos, who has secured the funding for the project, and Dr. Michael D. Naish, who has provided guidance for the development of the project. Five articles were chosen to form the main body of the thesis—four articles have been published in conference proceedings and peer reviewed journals, and one article is under review for publication. The extent of the collaboration of the co-authors is listed below.

Chapter 3. Tremor Characterization and Estimation

Paper: Characterization of Parkinsonian Hand Tremor and Validation of a High-Order Tremor Estimator

Current Status: Published in IEEE Transactions on Neural Systems and Rehabilitation Engineering, vol. 26, no. 9, September 2018

Yue Zhou: First Author, developed the experimental protocol, collected patient data, developed the tremor estimator, analyzed the data, evaluated the tremor estimator with collected patient data, and wrote the manuscript.

Dr. Mary E. Jenkins: Co-author, recruited 18 participants with PD, supervised the experimental trial, and assisted in the editing and correcting of the manuscript.

Dr. Michael D. Naish: Co-author, supervised the development of the experimental setup and the data analysis, and edited and corrected the manuscript.

Dr. Ana Luisa Trejos: Corresponding author, supervised the development of the experimental setup and the data analysis, and edited and corrected the manuscript.

Chapter 4. First Generation WTSG

Paper 1: Development and Validation of a Finger Tremor Simulator

Current Status: Published in the Proceedings of the IEEE Canadian Conference on Electrical & Computer Engineering, May 2018
Yue Zhou: First Author, developed the physical tremor simulator and the experimental evaluation setup, collected experimental data, analyzed the data, and wrote the manuscript.

Dr. Michael D. Naish: Co-author, supervised the development of the simulator and the experimental setup, supervised the data analysis, and edited and corrected the manuscript.

Dr. Mary E. Jenkins: Co-author, assisted in the editing and correction of the manuscript.

Dr. Ana Luisa Trejos: Corresponding author, supervised the development of the simulator and the experimental setup, supervised the data analysis, and edited and corrected the manuscript.

**Paper 2**: Development of a Wearable Tremor Suppression Glove

**Current Status**: Published in the Proceedings of the IEEE International Conference on Biomedical Robotics and Biomechatronics, October 2018

Yue Zhou: First Author, developed the first generation wearable tremor suppression glove and the experimental evaluation setup, collected experimental data, analyzed the data, and wrote the manuscript.

Dr. Mary E. Jenkins: Co-author, assisted in the editing and correcting of the manuscript.

Dr. Michael D. Naish: Co-author, supervised the development of the wearable tremor suppression glove and the experimental setup, supervised the data analysis, and edited and corrected the manuscript.

Dr. Ana Luisa Trejos: Corresponding author, supervised the development of the wearable tremor suppression glove and the experimental setup, supervised the data analysis, and edited and corrected the manuscript.

**Chapter 5. Second Generation WTSG and Pre-clinical Validation**

**Paper**: An Alternative Approach for Pathological Tremor Management – A Wearable Tremor Suppression Glove: Design and Validation

**Current Status**: Submitted to IEEE/ASME Transactions on Mechatronics, May 2019

Yue Zhou: First Author, developed the second generation wearable tremor suppression glove and the pre-clinical evaluation setup, developed the experimental protocol, collected the data from one subject with tremor wearing the glove, analyzed the data, and wrote the manuscript.

Anas Ibrahim: Co-author, assisted in the pre-clinical patient trial.
Dr. Kenneth G. Hardy: Co-author, participated the trial as patient collaborator

Dr. Mary E. Jenkins: Co-author, assisted in the editing and correcting of the manuscript.

Dr. Michael D. Naish: Co-author, supervised the development of the second generation wearable tremor suppression glove and the experimental setup, supervised the data analysis, and edited and corrected the manuscript.

Dr. Ana Luisa Trejos: Corresponding author, supervised the development of the second generation wearable tremor suppression glove and the experimental setup, supervised the data analysis, and edited and corrected the manuscript.

Chapter 6. Spatial Impact of Suppressed Tremor on Other Joints

Paper: Impact of Suppressed Tremor: is suppression of proximal joints sufficient?

Current Status: Published in the Proceedings of the IEEE EMBS International Conference on Biomedical & Health Informatics, March 2018

Yue Zhou: First Author, developed the experimental protocol, collected patient data, analyzed the data, and wrote the manuscript.

Dr. Mary E. Jenkins: Co-author, recruited 14 participants with PD, supervised the experimental trial, and assisted in the editing and correcting of the manuscript.

Dr. Michael D. Naish: Co-author, supervised the development of the experimental setup and the data analysis, and edited and corrected the manuscript.

Dr. Ana Luisa Trejos: Corresponding author, supervised the development of the experimental setup and the data analysis, and edited and corrected the manuscript.
Acknowledgements

It has been four years since I started my PhD program at Western. People say that time is relative; indeed, at the time I’m writing this sentence I still can’t realize that I’m close to the end of my study. I can still remember the very first day when I joined the WearME Lab—it feels like it happened yesterday. I guess I have enjoyed so much working with my supervisors, my colleagues and my friends, that I lost track of time. Just like everyone has to wake up from their dreams eventually, I’m finalizing the last chapter of my PhD study and preparing for the world with a mindset.

I consider myself lucky to have had the opportunity to continue this work with my supervisors, Dr. Ana Luisa Trejos and Dr. Michael Naish after I received my Master’s Degree. It was their insightful supervision, consistent support and constructive guidance that helped me to get this far. Without them I wouldn’t be where I am now. I would like to express my deepest gratitude to both of them.

This tremor project wouldn’t have made it this far without the consistent support and patient scheduling from Dr. Mary Jenkins. Although Dr. Jenkins isn’t my supervisor, she has provided me with extensive help and clinical knowledge. I have really enjoyed every trial we worked on together. My gratitude to you goes beyond what words can describe.

I would also like to thank my advisor Dr. Lyndon Brown. Your knowledge and suggestions have provided me with so many ideas in this project. I would really like to thank you for keeping me on track.

Anas and Memo, there are so many great ideas generated from our everyday Tim Hortons trip. People say their ideas come to them while they are in the shower. I’d say our ideas come to us on
the way to Tims. I have really enjoyed our conversations everyday and having you guys in the lab. Other than work related topics, our after 5 pm exercise always keeps me excited, not to mention the fishing trip, game day, grill day, etc. I’m really glad that I have met you guys during my study.

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Having some place to go is home; having someone to love is family; having both is blessing. My family has provided me with such blessing unconditionally since I was born to this world. I’m trying to find the most appropriate word to express what I feel in my heart, but at this moment, no words seem to be able to express how much I appreciate their love and support. I’m just happy and lucky to be part of our family.

I would like to leave my appreciation for my beloved wife Lule Aliraj to the very end of these Acknowledgments. Her consistent support, love and patience have given me courage and energy to face many difficulties during the past four years. Life is beautiful, exciting, difficult, and sometimes scary, but with you by my side, I see the life I dreamed of.
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Nomenclature and Acronyms

Latin Letters

$a$ Filter convergence coefficient
$a_{n1}$ Fourier coefficient of $S_{t1}$
$a_{n2}$ Fourier coefficient of $S_{t2}$
$a_{n3}$ Fourier coefficient of $S_{t3}$
$A$ Raw signal
$A_c$ Calibrated data
$A_g$ Earth gravity matrix
$A_L$ Linear acceleration caused directly by the tremor motion
$A_r$ Sampled data in line with the known Earth gravity matrix
$b_{n1}$ Fourier coefficient of $S_{t1}$
$b_{n2}$ Fourier coefficient of $S_{t2}$
$b_{n3}$ Fourier coefficient of $S_{t3}$
$CM$ Calibration matrix
$d$ Distance between the fixture and the joint
$f_{t1}$ The estimated frequency of the first tremor harmonic
$f_{t2}$ The estimated frequency of the second tremor harmonic
$f_{t3}$ The estimated frequency of the third tremor harmonic
$F$ State transition matrix of the Kalman Filter
NOMENCLATURE AND ACRONYMS

\( F_s \) Stall force
\( g \) Filtered gravity component
\( G \) Process noise covariance matrix of the Kalman Filter
\( H \) Measurement matrix of the Kalman Filter
\( i \) Current time step
\( k \) Harmonic number
\( K \) Kalman gain
\( M \) Number of harmonics
\( n \) The end sample of a signal with finite length
\( N \) Number of harmonics
\( p \) Amplitude weight
\( \mathbf{P} \) Vector of the amplitude weights
\( Q \) Covariance of the process noise
\( R \) Measurement noise covariance matrix of the Kalman Filter
\( S \) Innovation covariance of the Kalman Filter
\( S_f \) Band-pass filtered signal
\( S_{\text{rms}} \) RMS value of the tremor magnitude
\( S_t \) Tremor magnitude
\( S_x \) Amplitude of the tremor motion in the \( x \) direction
\( S_y \) Amplitude of the tremor motion in the \( y \) direction
\( S_z \) Amplitude of the tremor motion in the \( z \) direction
\( S_{t1} \) The first tremor harmonic
\( S_{t2} \) The second tremor harmonic
\( S_{t3} \) The third tremor harmonic
\( t \) Estimated tremor motion
\( T_o \) Elapsed CPU time of the estimator under comparison with the enhanced HWFLC-KF
NOMENCLATURE AND ACRONYMS

\( T_p \) Elapsed CPU time of the enhanced HWFLC-KF
\( T_s \) Time step
\( u \) Angular velocity of the joint measured from the IMUs
\( X \) State vector containing the time varying frequency weight of the estimated signal
\( y \) Estimated voluntary velocity signal

Greek Letters

\( \epsilon \) Estimation error of the WFLC
\( \omega \) Frequency weight
\( \mu \) Amplitude adaptation gain
\( \nu \) Frequency adaptation gain
\( \pi \) Pi
\( \tau_t \) Tremor torque
\( \theta \) Joint angle

Acronyms

ADL Activities of Daily Living
AFFE Activation of the Forearm Flexors and Extensors
ANOVA Analysis of Variance
API Application Program Interface
BLDC Brushless DC
BMFLC Bandlimited Multiple Fourier Linear Combiner
BPF Band-Pass Filter
CAD Computer-Aided Design
DC Direct Current
DIP Distal Interphalangeal
<table>
<thead>
<tr>
<th>Acronym</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>DOF</td>
<td>Degrees of Freedom</td>
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<tr>
<td>DRIFT</td>
<td>Dynamically Responsive Intervention for Tremor Suppression</td>
</tr>
<tr>
<td>EFE</td>
<td>Elbow Flexion–Extension</td>
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<tr>
<td>EM</td>
<td>Electromagnetic</td>
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<td>EMG</td>
<td>Electromyography</td>
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<tr>
<td>ET</td>
<td>Essential Tremor</td>
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<tr>
<td>FES</td>
<td>Functional Electrical Stimulation</td>
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<tr>
<td>FFT</td>
<td>Fast Fourier Transform</td>
</tr>
<tr>
<td>FLC</td>
<td>Fourier Linear Combiner</td>
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<tr>
<td>FPS</td>
<td>Forearm Pronation–Supination</td>
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<td>HWFLC-KF</td>
<td>High-Order WFLC-based Kalman Filter</td>
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<td>IF</td>
<td>Index Finger</td>
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<td>IMU</td>
<td>Inertial Measurement Unit</td>
</tr>
<tr>
<td>KF</td>
<td>Kalman Filter</td>
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<tr>
<td>LMS</td>
<td>Least Mean Square</td>
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<td>MCP</td>
<td>Metacarpophalangeal</td>
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<td>MDS</td>
<td>Movement Disorder Society</td>
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<td>MDS-UPDRS</td>
<td>MDS-Sponsored Revision of the Unified Parkinson’s Disease Rating Scale</td>
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<tr>
<td>MMS</td>
<td>Multi-Channel Mechatronic Splitter</td>
</tr>
<tr>
<td>MRD</td>
<td>Magnetorheological Damper</td>
</tr>
<tr>
<td>MRF</td>
<td>Magnetorheological Fluid</td>
</tr>
<tr>
<td>PC</td>
<td>Personal Computer</td>
</tr>
<tr>
<td>PD</td>
<td>Parkinson’s Disease</td>
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<tr>
<td>PEA</td>
<td>Percentage Estimation Accuracy</td>
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<tr>
<td>PID</td>
<td>Proportional–Integral–Derivative</td>
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<td>PIP</td>
<td>Proximal Interphalangeal</td>
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<tr>
<td>Acronym</td>
<td>Definition</td>
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<tr>
<td>PSD</td>
<td>Power Spectral Density</td>
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<tr>
<td>RMS</td>
<td>Root Mean Square</td>
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<tr>
<td>RMSE</td>
<td>Root Mean Square Error</td>
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<tr>
<td>SD</td>
<td>Standard Deviation</td>
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<tr>
<td>SPI</td>
<td>Serial Peripheral Interface</td>
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<td>T</td>
<td>Thumb</td>
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<tr>
<td>W</td>
<td>Wrist</td>
</tr>
<tr>
<td>WD</td>
<td>Wrist Radial Deviation</td>
</tr>
<tr>
<td>WFE</td>
<td>Wrist Flexion-Extension</td>
</tr>
<tr>
<td>WFLC</td>
<td>Weighted-Frequency Fourier Linear Combiner</td>
</tr>
<tr>
<td>WOTAS</td>
<td>Wearable Orthosis for Tremor Assessment and Suppression</td>
</tr>
<tr>
<td>WTSG</td>
<td>Wearable Tremor Suppression Glove</td>
</tr>
<tr>
<td>WTSD</td>
<td>Wearable Tremor Suppression Device</td>
</tr>
<tr>
<td>3D</td>
<td>Three-Dimensional</td>
</tr>
</tbody>
</table>
Chapter 1

Introduction

Parkinson’s disease (PD) has been recognized as the second most prevalent chronic neurological disease after Alzheimer’s disease. The cause of PD is characterized by dopaminergic-cell loss in the basal ganglia, the internal segment of the globus pallidus, and the pars reticulata of the substantia nigra [1–3]. Recent studies in the field of the epidemiology of PD [4–8] have estimated that the incidences of PD in Canada, China, UK, USA, and Australia range from 16 to 450 per 100,000 persons and the overall number has been predicted to double in the next 15 years [8–11]. While the increased life span afforded by modern medicine is appreciated, knowledge of the possible higher incidence of PD may make one’s retirement life less enjoyable than it should be, and the progressive and chronic nature of PD is expected to lead to a substantial economic burden on affected individuals and the health care system [10, 12–15]. Understanding the disease and finding additional treatment options for PD are crucial to both the affected individuals and society. Among the symptoms of PD, tremor significantly reduces the quality of life, creating difficulties in many activities of daily living (ADL), such as writing and eating. Furthermore, the social embarrassment caused by tremor makes the affected individuals avoid social contact outside of their own family and friends.
1.1 Motivation

Parkinsonian tremor often appears as a pill-rolling action of the hand. It appears as an unintentional and oscillatory movement with a time-varying magnitude and relatively stable frequency [16] in the upper limb. There are two types of parkinsonian tremor defined clinically in people with PD, one is resting tremor, which appears when no muscle contractions are made, and the other one is action tremor, which appears during muscle contraction [17].

The current treatments for tremor have varying degrees of effectiveness and often are associated with side effects. Medication, being the mainstay of treatment, is usually less effective for treating tremor than rigidity and bradykinesia, and often presents side effects, such as nausea, depression, headache, anxiety, and issues with drug tolerance [18, 19]. Brain surgeries, such as deep brain stimulation and stereotactic thalamotomy [20, 21], being the second option of treatment, are often very effective. However, the invasive nature of these treatments limits their use to only the cases of severe tremor. Moreover, they carry a significant risk of complications for the individual [22], such as infection, or nerve damage. Given the side effects and complications of traditional treatments, the emergence of wearable technology facilitates an externally worn mechatronic device as a potential alternative approach for managing tremor.

Adelstein et al. discovered that tremor can be effectively suppressed by applying mechanical loading [23], which has encouraged researchers to develop externally worn exoskeleton-type mechatronic devices for tremor management [24–40]. Although these devices have been shown to suppress tremor, most of them have not been validated on individuals with PD due to the issues associated with their size, weight, power source, and overall effectiveness. Furthermore, hand tremor has been neglected by most researchers. Thus, there is a need to study the nature of hand tremor, to develop a lightweight and compact wearable tremor suppression device (WTSD) for the treatment of hand tremor, and to validate it on a human subject with parkinsonian tremor.

1.2 General Problem Statement

Ideally, with the use of a fully developed WTSD, an individual with PD could perform ADL with less effort and stress. Unfortunately, none of the developed WTSDs have become commercially
available, and the validation of most devices stopped at either simulation, or with healthy volunteers. In addition, there is a large proportion of individuals with PD (around 25%) in whom traditional treatments do not achieve a significant reduction in tremor [41]. In order to provide an alternative solution for the afflicted population, there is a need to promote the translation of a WTSD from the lab to the clinic, and eventually to commercial use.

Nevertheless, tremor suppression in the elbow and wrist have been studied and reported extensively; however, fewer studies have addressed tremor in the fingers. Thus, there is a need to fill the knowledge gap by studying the nature of hand tremor and to develop a compact and lightweight device that is capable of suppressing hand tremor.

This work proposes to develop a wearable tremor suppression glove (WTSG) that is capable of suppressing tremor in the fingers and the wrist while permitting voluntary motion to occur, to validate the feasibility of the WTSG on an individual with PD, and to acquire comprehensive perspectives on the WTSG from the user.

1.3 Research Objectives

The overall goals of this thesis are to study parkinsonian hand tremor, and to develop and validate a WTSG that is capable of suppressing parkinsonian hand tremor while allowing the user’s voluntary motion. To achieve these goals, the work has focused on the following specific objectives:

1. To conduct a detailed analysis of collected hand tremor data to quantify and extract the features of parkinsonian hand tremor.
   (a) Extract the bio-mechanical and bio-electrical features of hand tremor.
   (b) Study the impact of tremor from the suppressed joint(s) on the other joint(s).
   (c) Study the bio-mechanical correlation in each joint under the impact of tremor.

2. To develop a control algorithm that allows the actuation system to maximize the suppression of tremor motion while minimizing the influence on the voluntary motion of the user.

3. To develop and validate the design of a WTSG.
(a) Develop a WTSG that utilizes mechanical loading to suppress hand tremor.

(b) Validate the WTSG on a physical tremor simulator.

(c) Validate the WTSG on an individual with PD in a pre-clinical trial.

1.4 Thesis Outline and Contributions

This section summarizes the structure of the rest of the thesis and the main contributions of each chapter as follows:

Chapter 2  
Literature Review: Presents a comprehensive review of the characteristics of parkinsonian tremor, the state of the art in tremor estimation techniques, and existing tremor suppression devices.

Chapter 3  
Tremor Characterization and Estimation: Presents the experimental setups and protocols for tremor data collection, the analysis of the data collected, and the development and validation of an enhanced high-order tremor estimator. The objectives of this study were to further our understanding of the quantitative features of parkinsonian tremor in the fingers and wrist and to develop a novel tremor estimator based on the features extracted from the data collection. The contributions of this study include the following:

• The kinematic and electro-physiological characteristics of parkinsonian hand tremor in both the time domain and the frequency domain were quantified. The joints of interest include the index finger, thumb metacarpophalangeal (MCP) joints, and wrist.

• Frequency analysis of the tremor data showed a distinct pattern that has been neglected by other researchers, namely that parkinsonian hand tremor consists of multiple harmonics. The frequency of the $k^{th}$ harmonic is roughly $k$ times the fundamental frequency.

• An enhanced high-order Weighted-Frequency Linear Combiner-based Kalman Filter was proposed and evaluated for the purpose of tremor estimation. The per-
1.4 Thesis Outline and Contributions

centage estimation accuracy achieved from the proposed estimator is $96.3 \pm 1.7\%$, with average improvements of $28.5\%$ and $48.9\%$ over its lower-order counterpart and the Weighted Fourier Linear Combiner.

Chapter 4

First Generation WTSG: Outlines the development of the first prototype of the WTSG and the experimental validation on a physical tremor simulator. The objectives of this study were to develop and validate a WTSG that can be used to mechanically suppress parkinsonian tremor in the hand without affecting the voluntary motion. The contributions of this study are the following:

- A physical tremor simulator was developed and evaluated for the validation of a WTSG. This simulator can reproduce recorded tremor motion, tremor torque, and voluntary motion.

- A WTSG prototype was designed and validated on the physical tremor simulator using recorded tremor data from seven subjects with PD. This WTSG suppresses tremor in the index finger MCP joint, thumb MCP joint, and wrist in the direction of flexion and extension. Two assessments were performed to evaluate the performance of tremor suppression and voluntary motion following. The tremor suppression assessment showed an overall tremor amplitude reduction of $85.0\% \pm 8.1\%$, and the voluntary motion following assessment showed a root mean square error (RMSE) of $14.2\% \pm 2.5\%$ and a correlation coefficient of $0.97 \pm 0.01$.

Chapter 5

Second Generation WTSG and Pre-clinical Validation: Presents an improved version of the original prototype, as well as the validation of its performance on an individual with PD. The objectives of this study are to improve the design of the WTSG based on the performance evaluation of the first generation, and to validate it on a human subject with parkinsonian tremor. The contributions of this study are as follow:

- The mechanical design of the WTSG was improved to address the issues identified in the first generation WTSG.
• A pre-clinical trial was performed on one subject with parkinsonian tremor to validate the feasibility of using the WTSG to suppress human tremor. Results from the experiment showed an overall suppression of 73.1%, 80.7% and 85.5% in resting tremor, 70.2%, 79.5% and 81% in postural tremor, and 52.6%, 55.5% and 59.8% in kinetic tremor in the index finger MCP joint, the thumb MCP joint and the wrist, respectively.

Chapter 6 Spatial Impact of Suppressed Tremor: Presents a study on the impact of suppressed tremor on the other unrestricted joints. The objectives of this study were to investigate how suppressed tremor influences the tremor in the unrestricted joints, and to determine the importance of suppressing finger tremor in addition to the other most-studied joints. The contributions of this study are as follow:

• The kinematic magnitude of the tremor in the unrestricted joints was increased when the tremor in the other joints was suppressed. This increase was particularly significant in the distal joints when the proximal joints were suppressed. This indicates that it is not enough to suppress only the elbow and wrist tremor—the suppression of finger tremor should be considered as well.

• The analysis of the tremor EMG signal did not show a significant change, which indicates that the increase in the magnitude of tremor motion may be the result of the propagation of tremor from one joint to another.

Chapter 7 Conclusions and Future Work: Presents the conclusions of the entire thesis and proposes suggestions for the future work.
Bibliography


Chapter 2

Literature Review

2.1 Introduction

In order to establish the design requirements of a WTSG, the characteristics of parkinsonian tremor, existing tremor estimation techniques, and available tremor suppression devices are presented in this chapter. Several comprehensive literature reviews were performed during the periods of January to June 2016, September to December 2017, June to September 2018, and July 2019 using Google Scholar search engine with the following keywords: Parkinson’s disease, tremor, tremor characterization, tremor estimation, tremor suppression device, assistive device and tremor simulator. These literature review sessions contributed the works presented in Chapter 3 to Chapter 6 of this thesis. The paper inclusion criteria limits the reviewing priority to the papers published in the previous 15 years. However, the papers published earlier are included if they deemed to be important to the narrative of the review. As a result, a total of 90 papers were added in the reference list.

Based on the reviewed papers, the remainder of this chapter is organized as follows: Section 2.2 reviews the characterization of parkinsonian tremor, Section 2.3 reviews tremor estimation techniques, Section 2.4 reviews the existing tremor suppression devices, which have been divided into three categories according to the type of stimulation used for suppressing tremor, and finally, Section 2.5 reviews the tremor simulation devices.
2.2 Characterization of Parkinsonian Tremor

In general, tremor can be categorized into physiological tremor and pathological tremor. Physiological tremor is often a benign inherent component of voluntary movement. Since its magnitude is negligible compared to the magnitude of voluntary movement, the suppression of physiological tremor is commonly seen in the tasks that require precise movement, such as those performed during microsurgery [1]. Pathological tremor, in contrast, significantly affects the mobility of affected individuals.

Parkinsonian tremor, as one of the common types of pathological tremor, consists of resting tremor that occurs when no muscle contractions are made, and postural tremor that occurs during muscle contraction [2–4]. The reported frequency range of resting tremor in the upper limb is 3–17.3 Hz, and 3.9–16.8 Hz for postural tremor [2–6]. Although some studies have reported that parkinsonian tremor only consists of single harmonic [7–10], recent evidence has demonstrated that parkinsonian tremor consists of multiple harmonics [6, 11, 12]. The frequency of the $k^{th}$ harmonic is $k$ times the fundamental frequency. As a summary, the magnitudes of the bio-mechanical features of the resting tremor and the postural tremor in the upper-limb joints are shown in Table 2.1 and Table 2.2.

2.3 Tremor Estimation Techniques

One of the major steps towards the suppression of tremor using a wearable tremor suppression device is to estimate the tremor component in the presence of voluntary motion. Since the frequency of tremor is typically higher than the frequency of voluntary motion, classic filters, such as low-pass filters [16–18], high-pass filters [19, 20] and notch filters [21, 22], were widely adopted by most of the studies in the field to prove the feasibility of suppressing tremor using mechatronic devices. However, the drawbacks of these filters include the inherent phase delay and amplitude attenuation.

To reduce the estimation error caused by these factors, a number of techniques have been proposed to minimize the phase lag and amplitude attenuation. For example, a pair of cascaded low-pass and high-pass Infinite Impulse Response filters [23] were proposed that compensate for
Table 2.1: RMS Magnitude of Parkinsonian Resting Tremor in the Direction of Flexion–Extension

<table>
<thead>
<tr>
<th></th>
<th>Elbow</th>
<th>Wrist</th>
<th>IF MCP(^1)</th>
<th>T MCP(^1)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Linear Acceleration (m/s(^2))</strong></td>
<td>0.39 [13]</td>
<td>0.9 ± 0.5 [6]</td>
<td>2.1 ± 1.2 [6]</td>
<td>1.8 ± 1.1 [6]</td>
</tr>
<tr>
<td><strong>RMS Angular Displacement (degrees)</strong></td>
<td>/</td>
<td>/</td>
<td>2.9 ± 1.8 [6]</td>
<td>2.2 ± 1.6 [6]</td>
</tr>
<tr>
<td><strong>Torque (Nm)</strong></td>
<td>0.1 [15]</td>
<td>0.02 [15]</td>
<td>/</td>
<td>/</td>
</tr>
</tbody>
</table>

\(^1\) IF MCP - Index finger metacarpophalangeal joint

2 T MCP - Thumb metacarpophalangeal joint

Table 2.2: RMS Magnitude of Parkinsonian Postural Tremor in the Direction of Flexion–Extension

<table>
<thead>
<tr>
<th></th>
<th>Elbow</th>
<th>Wrist</th>
<th>IF MCP</th>
<th>T MCP</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Linear Acceleration (m/s(^2))</strong></td>
<td>&lt;5 [13]</td>
<td>1.5 ± 0.8 [6]</td>
<td>2.7 ± 1.0 [6]</td>
<td>2.5 ± 1.3 [6]</td>
</tr>
<tr>
<td><strong>RMS Angular Velocity (degrees/s)</strong></td>
<td>/</td>
<td>48.8 ± 23.2 [6]</td>
<td>77.7 ± 38.2 [6]</td>
<td>64.0 ± 32.5 [6]</td>
</tr>
<tr>
<td><strong>RMS Angular Displacement (degrees)</strong></td>
<td>/</td>
<td>/</td>
<td>3.1 ± 1.3 [6]</td>
<td>2.4 ± 1.2 [6]</td>
</tr>
<tr>
<td><strong>Torque (Nm)</strong></td>
<td>1.2 [14]</td>
<td>0.2 [14]</td>
<td>/</td>
<td>/</td>
</tr>
</tbody>
</table>

the phase lag caused by the low-pass filter by introducing a phase lead through the high-pass filter. In another example, a backstepping-sliding mode control algorithm [24] was proposed to reduce the phase lag between the resulted tremor and the corresponding muscle torque by estimating the muscle torque. Finally, an adaptive band-pass filter [25] that reduces the phase lag by updating the center frequency of the filter according to the input signals was also suggested. Since the tremor motion is often at a higher frequency than the voluntary motion, a notch filter has been used to suppress those signals at tremor frequency rather than all frequencies beyond a certain point [8, 22, 26, 27]. This results in less distortion to the filtered signal. This type of filter works well on signals with fixed frequency, however, the frequency of tremor is time-varying, and therefore the notch filter with fixed frequency cannot capture the tremor motion without losing features.

Considering the drawbacks of the aforementioned filters, it is important to include adaptability to the design of a tremor estimator. Riviere et al. [28] proposed a Weighted-frequency Fourier Linear Combiner (WFLC), which functions as an adaptive notch filter that adaptively controls
the notch depth and notch frequency according to the input tremor signal. Specifically, this estimator was developed based on a Fourier Linear Combiner [29] with a modified tremor frequency estimation algorithm. The adaptation used in the WFLC for updating the tremor frequency and the tremor amplitude was accomplished using the least mean squares (LMS) algorithm. Given that Riviere’s WFLC adaptively controls only the notch depth and notch frequency, Nho [30] discovered that it is important to also consider controlling the notch bandwidth. Although the use of the gradient descent algorithm allows the WFLC to adapt to the frequency and amplitude of tremor with zero-phase shift [31], it can only adapt to a single-harmonic signal without degrading its performance [32, 33].

To overcome the drawbacks of the WFLC, Veluvolu et al. proposed a Bandlimited Multiple Fourier Linear combiner (BMFLC) [34]. The principle of the BMFLC for estimating signals with multiple frequencies is by incorporating multiple FLCs. Each FLC estimates a certain frequency band, and the combination of all FLCs produces a reconstructed signal of the signal of interest. However, the main drawback of BMFLC is that it requires prior knowledge of the frequency, and it works only on signals with fixed frequencies. Upon the realization of the drawbacks of BMFLC, Veluvolu et al. proposed a double adaptive BMFLC [35], and Wang et al. proposed an adaptive sliding BMFLC [36]. Both modified BMFLCs possess the advantages of BMFLC, and incorporate adaptive methods for the selection of the frequency band.

The WFLC-based estimators extract the tremor signal based on the estimation of the gradient of the mean square error using a truncated Fourier series with prior knowledge of the input signal, i.e., tremor. In contrast, the Kalman filter (KF) does not require any a priori assumption and it computes the optimal solution by minimizing the covariance of the a posteriori estimation error [37]. In addition, the KF is considered to be more robust for nonperiodic motions, such as parkinsonian tremor. Recent studies on using the KF and its derivative algorithms [12, 33, 38] have shown better accuracy in tremor estimation than the gradient decent methods.

Lastly, other than the aforementioned widely adopted tremor estimators, a filtered-X least mean square algorithm [39] was designed to estimate tremor. This algorithm works well on single-frequency tremor. However, it still requires further evaluation on real tremor signals. In addition, Zhang et al. [40] proposed a tremor estimator that incorporates both Empirical Mode Decompo-
sition and the Hilbert-Huang Transform. Comparing it to the aforementioned approaches, this estimator does not lose features by approximating tremor as a stationary process, since Empirical Mode Decomposition can decompose the nonlinear and nonstationary tremor signal into a series of stationary signals, which can then be used in real-time tremor detection.

2.4 Tremor Suppression Devices: State of the Art

Tremor estimator separates the tremor motion signal from the voluntary motion signal. To suppress the tremor motion, a number of approaches have been developed over the past a few decades. Adelstein et al. [41] discovered that tremor can be effectively suppressed by applying mechanical loading, which has encouraged researchers to develop externally worn exoskeleton-type mechatronic devices for tremor management. As the technology evolves, more suppression methods have been established. These methods can be classified into four categories based on the suppression modality, i.e., mechanical loading, electrical stimulation, afferent pathway stimulation and mechanical vibration. The evolution of the WTSDs developed with these modalities is illustrated in Fig. 2.1 in chronological order.

2.4.1 Mechanical Loading-based Tremor Suppression Devices

Mechanical loading-based tremor suppression technology, being the most popular and studied technology, can be classified by the suppression type, i.e., active suppression [8, 11, 17, 19, 24, 42–63], semi-active suppression [20, 64–67] and passive suppression [68–70]. The features of these devices are listed in Table 2.3.

Suppressing tremor using active suppression device implies an active interaction applied by the device on the user. These devices often use actuators that can produce motion and force by electric current, pneumatic pressure and hydraulic fluid pressure. Pons et al. [8, 50–57, 63] developed a three degree of freedom (DOF) Wearable Orthosis for Tremor Assessment and Suppression (WOTAS). This device employs both active and passive suppression strategies using electric motors to suppress tremor with minimal impact on the user’s voluntary motion. WOTAS supports tremor suppression in the directions of elbow flexion–extension, forearm adduction–abduction, and wrist
Table 2.3: Existing wearable tremor suppression devices using mechanical loading method. Figures used with permission.

<table>
<thead>
<tr>
<th>Device</th>
<th>Suppression Type</th>
<th>Suppression Mechanism</th>
<th>DOF</th>
<th>Validation Method</th>
<th>Weight (kg)</th>
<th>Suppression Rate</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Active</td>
<td>DC motor</td>
<td>EFE</td>
<td>10 tremor subjects</td>
<td>0.85</td>
<td>40%</td>
<td>Room et al. [8, 50–57, 63]</td>
</tr>
<tr>
<td></td>
<td>Active</td>
<td>DC motor</td>
<td>EFE</td>
<td>N.A.</td>
<td>0.33</td>
<td>N.A.</td>
<td>Seki et al. [17, 45–47]</td>
</tr>
<tr>
<td></td>
<td>Active</td>
<td>DC motor</td>
<td>EFE</td>
<td>one tremor subject</td>
<td>0.41</td>
<td>50–80%</td>
<td>Matsumoto et al. [48, 49]</td>
</tr>
<tr>
<td></td>
<td>Active</td>
<td>Pneumatic cylinder</td>
<td>WFE, FPS, EFE, WD</td>
<td>10 tremor datasets</td>
<td>0.378</td>
<td>98.8%</td>
<td>Taheri et al. [11, 19, 24, 61, 62]</td>
</tr>
</tbody>
</table>
### 2.4 Tremor Suppression Devices: State of the Art

<table>
<thead>
<tr>
<th>Reference</th>
<th>DOF</th>
<th>Suppression Mechanism</th>
<th>Suppression Type</th>
<th>Suppression Rate</th>
<th>Weight (kg)</th>
<th>Validation Method</th>
</tr>
</thead>
<tbody>
<tr>
<td>Herrnstadt et al. [42–44]</td>
<td>EFE</td>
<td>DC motor</td>
<td>Active</td>
<td>99.8%</td>
<td>0.875</td>
<td>One tremor dataset</td>
</tr>
<tr>
<td>Huen et al. [58]</td>
<td>WFE,</td>
<td>DC motor</td>
<td>Active</td>
<td>77%</td>
<td>0.35</td>
<td>Six healthy subjects with simulated tremor</td>
</tr>
<tr>
<td>Zhou et al. [59]</td>
<td>WFE</td>
<td>DC motor</td>
<td>Active</td>
<td>N.A.</td>
<td>0.229</td>
<td></td>
</tr>
<tr>
<td>Kotovsky et al. [68]</td>
<td>WFE</td>
<td>viscous beam</td>
<td>Passive</td>
<td>N.A.</td>
<td>0.265</td>
<td>Five tremor subjects</td>
</tr>
</tbody>
</table>

- **Device**: Images of the suppression devices are provided.
### 2.4 Tremor Suppression Devices: State of the Art

<table>
<thead>
<tr>
<th>Device Type</th>
<th>Suppression Mechanism</th>
<th>DOF</th>
<th>Suppression Rate</th>
<th>Validation Method</th>
<th>Weight (kg)</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Passive</td>
<td>Tuned Mass FPS</td>
<td>One</td>
<td>20–82%</td>
<td>One healthy subject with simulated tremor</td>
<td>0.28</td>
<td>Takanokura et al. [69]</td>
</tr>
<tr>
<td>Semi-active</td>
<td>Double viscous beam</td>
<td>FPS</td>
<td>98%</td>
<td>One tremor subject</td>
<td>0.2</td>
<td>Loureiro et al. [67]</td>
</tr>
<tr>
<td>Semi-active</td>
<td>Double viscous beam</td>
<td>EFE</td>
<td>88%</td>
<td>Three healthy subjects with simulated tremor</td>
<td>0.942</td>
<td>Herrnstadt et al. [20]</td>
</tr>
</tbody>
</table>
2.4 Tremor Suppression Devices: State of the Art

Figure 2.1: Development of upper limb wearable tremor suppression devices. ML: Mechanical Loading; FES: Functional Electrical Stimulation; APS: Afferent Pathway Stimulation; MV: Mechanical Vibration.

Fijie et al. [17, 45–47] developed an active elbow tremor suppression exoskeleton device that suppresses tremor using an electric motor. This device estimates elbow tremor in the direction of flexion–extension using the EMG signal acquired from the user’s biceps and triceps. The total weight of the device is 0.33 kg. Although the device has been developed, no efficacy study has been conducted to show the performance of this device when suppressing tremor. Following the development of this device, a second prototype [48, 49] was developed by the same research group to improve the wearability of the first prototype. The change of the mounting mechanism of the device resulted in an increase in the total weight to 0.41 kg. This prototype was tested on one subject with essential tremor (ET). The result showed a 50–80% reduction in the tremor amplitude.

Richer et al. [11, 19, 24, 61, 62] investigated the use of pneumatic cylinders for suppressing wrist tremor in the directions of flexion–extension and radial–ulnar deviation. The control system
was tested on a benchtop setup with 10 recorded tremor datasets. The result showed an average of 98.8% reduction in tremor power. Similarly, Kalaivarasi et al. [66] introduced a two DOF tremor suppression device assisted with a pneumatic mechanism. In comparison with the pneumatic tremor suppression device developed by Richer’s group, only 30% suppression in tremor amplitude was obtained in this study. Zamanian and Richer [60] also proposed another tremor suppression device with the use of a permanent magnet linear motor. This device is lighter in weight (0.315 kg), and it does not require a compressed air source. The benchtop validation showed an average of 97.6% reduction in tremor power.

Herrnstadt and Menon [42–44] developed an active elbow tremor suppression orthosis that suppresses elbow tremor using electric motors with a gear transmission system. This device was tested on a benchtop setup with a recorded elbow tremor signal in the direction of flexion–extension. The result showed a 99.8% reduction in the tremor power. The total weight of the device is 0.875 kg.

Huen et al. [58] proposed a two DOF forearm robotic device for suppressing tremor in the directions of forearm pronation–supination and wrist flexion–extension. This device was validated on six healthy volunteers with simulated tremor motion generated by mechanical vibration. The result showed an average of 77% reduction in tremor amplitude. The total weight of the device including the electric motors is 0.35 kg.

Lastly, Zhou et al. [59] proposed a multi-channel mechatronic splitter (MMS) towards the development of a WTSD. This mechatronic system allows a single driving motor to support multiple independent outputs. The benchtop validation with seven parkinsonian tremor datasets showed that the MMS can track user’s voluntary motion with an average of 12.4% root mean square (RMS) error. The total weight of an MMS is 0.229 kg.

In comparison with the active suppression devices, passive suppression devices employ constant damping to the target tremor. Such approach allows these devices to suppress tremor with only mechanical damper(s). Therefore, the sensing system, control system and battery system are not required as the core components in a passive suppression device. Kotovsky et al. [68] developed a passive tremor suppression device using a viscous beam. This device was designed to suppress wrist tremor in the direction of flexion–extension. Although the validation of this device was not
2.4 Tremor Suppression Devices: State of the Art

Presented, it is clear that such device can reduce the amplitude of tremor. However, the same amount of damping is also expected to be applied to the voluntary motion of the user. Such drawback is the major obstacle for the use of passive suppression devices in a patient population. Similarly, Takanokura et al. [69] developed a three DOF passive tremor suppression device for managing tremor in the elbow (flexion–extension) and wrist (flexion–extension and ulnar–radial deviation). The validation on one healthy volunteer with simulated tremor showed 20–62% tremor reduction in the wrist and 82% in the elbow. Lastly, a similar device was developed by Katz et al. [70]. This device included the suppression in the direction of forearm pronation–supination. The benchtop validation achieved 86% tremor reduction with the proposed device.

Lastly, considering the drawbacks of the passive tremor suppression strategy, a semi-active tremor suppression strategy was proposed to reduce the impact on voluntary motion by employing controllable damping to the target joint. Since this strategy does not generate motion to the target joint, it is considered safer for using on humans compared to the active strategy. Richer et al. [64, 65] developed a semi-active tremor suppression device using magnetorheological dampers (MRDs). This MRD compensates the drawbacks of conventional dampers by using magnetorheological fluids (MRF). The viscosity of MRFs can be controlled by the intensity of the applied magnetic field. The concept of the proposed device includes the suppression in elbow flexion–extension, forearm pronation–supination, wrist ulnar–radial deviation and wrist flexion–extension. The benchtop validation with 10 tremor datasets showed an average of 96.3% and 74.3% reduction in tremor’s first and second harmonics, respectively. Two similar devices were found using the same magnetorheological technology. Loureiro et al. [67] tested the feasibility of the proposed device on one subject with tremor. This study achieved 98% tremor suppression rate in the direction of wrist flexion–extension. In addition, Hernstadt et al. [20] validated a semi-active elbow suppression brace on three healthy volunteers with simulated tremor. The result showed an average 88% reduction in tremor power.

2.4.2 Electrical Stimulation-based Tremor Suppression Devices

Devices using mechanical loading have been proven to have high suppression ratios, and when good models are available, are relatively easy to control. However, their size, weight, and power
consumption limit their clinical use. As an alternative, functional electrical stimulation (FES) has been proposed and studied as a way to suppress tremor. This approach uses modulated electrical signals generated by a functional electrical stimulator to evoke muscle contraction to reduce tremor. Depending on the method of delivering electrical stimulation to the target muscles, the FES-based tremor suppression devices can be classified into two categories: co-contraction stimulation [71–76] and out-of-phase stimulation [21, 77–81].

The co-contraction strategy generates continuous stimulation to both antagonist muscles in order to increase the stiffness of the target joint, and thereby reduces the amplitude of tremor motion. Grimaldi et al. [75] tested the co-contraction strategy on three subjects with tremor. Four pairs of FES electrodes were fixed on their flexor radialis carpi, extensor radialis carpi, biceps and triceps. Their tremor amplitude was measured by inertial measurement units (IMUs) on the index finger, wrist and elbow. The pulse frequency and pulse width were set to constant values of 30 Hz and 100 µs, while the pulse intensity varied from 10 to 34 mA. During the trial, each co-contraction pulse train followed a defined pattern, i.e., one second ramp increase, 20 second plateau, and one second ramp decrease. The measured tremor suppression ratio across all subjects was found to be 9% ± 35%. Although the suppression ratio is less than ideal, further investigation in optimizing suppression parameters and patterns may contribute to improving the suppression performance.

Gallego et al. [71, 72] implemented a neuroprosthesis that suppresses tremor in the wrist by regulating the level of muscle co-contraction. The level of muscle co-contraction was continuously adapted to the amplitude and frequency of the measured tremor from IMUs. The experimental validation involved six subjects with tremor. The results showed that the tremor amplitude was reduced by 52% ± 25%. Through the experiment with FES, the authors found that the etiology and symptomatology were not factors limiting the suppression performance of using FES-based system.

Bo et al. [73, 74] evaluated a new co-contraction stimulating strategy, in which the stimulation was intermittently turned on and off while the subjects were performing the task. The evaluation results on ten subjects indicate that eight out of ten subjects had consistent tremor suppression levels, with an average suppression ratio of 60% ± 27%. However, such reduction was not instan-
2.4 Tremor Suppression Devices: State of the Art

taneous for all trials, suggesting that prior training with FES may improve the overall suppression performance.

Lastly, a study that involved 34 subjects with resting tremor was conducted by Jitkritsadakul et al. [76]. This study implemented a constant pulse frequency of 50 Hz, a constant pulse width of 150 $\mu$s, and a constant pulse intensity of less than 20 mA over the abductor pollicis brevis and interosseus muscles of all subjects for 10 s. The tremor motion magnitude before and during stimulation were collected to quantify the suppression performance of the proposed FES system. The average tremor suppression ratio was 44% (SD=33%), and 21 subjects achieved at least a 30% tremor suppression level.

Comparing to the co-contraction strategy, the out-of-phase strategy applies electrical stimulation to the antagonistic muscles so that they generate force to compensate for the force generated by the tremorogenic muscles. Prochazka et al. [21] and Javidan et al. [77] developed and evaluated an FES-based tremor suppression system on 13 subjects with tremor. The participants’ triceps and biceps brachii were selected for stimulation. The effectiveness of tremor attenuation was assessed by comparing the measured wrist torque and acceleration with and without stimulation. This study showed a 53% ± 25% reduction of the tremor amplitude. No functional measures nor longer-term effects were assessed in this study. Most of the participants reported that the electrical stimulation was acceptable to continue with the experiment, except for one that found that the stimulation was painful. A further study [79] from the same research group compared the performance of FES on tremor suppression using digital and analog controllers. A total of three subjects with parkinsonian tremor were recruited to validate the proposed system. The experimental evaluation showed that the suppression ratio obtained from the analog controller was 69%, while the digital controller was 83%.

Different from previous work, Widjaja et al. [81] developed an FES-based tremor suppression system incorporating both electromyography (EMG) and acceleration as the input signals. The advantage of this system is that the EMG signal precedes the joint motion by 20–100 ms, which allows the system to detect the tremor in advance, thereby providing sufficient time for computation and muscle stimulation. This system was validated on the wrist flexors and extensors of one subject with ET. The pulse width, pulse frequency and pulse intensity of the FES system were
set constant to 200 µs, 25 Hz and 23 mA. The obtained suppression ratio was 57% at best, which is considerably low compared with the previous study. This is likely because a very simple on–off control was used in the system. The authors suggested that a more sophisticated controller may improve the suppression ratio of the proposed system.

The use of EMG signal in a tremor suppression device allows the control system to act on the target muscle/joint before the motion initiated, such advantage is the product of the electromechanical delay of the human musculoskeletal system. However, it is still a technical difficulty to process and to extract meaningful information from EMG signals. Maneski et al. [80] proposed a new FES hardware and software system designed to suppress tremor in the fingers, wrist, elbow and shoulder with inertial sensing feedback. In this study, a trapezoidal stimulation burst method was used to simulate a gradual activation profile, which is similar to the biological activation profile. This system was tested on a total of seven subjects with tremor, the pulse width and pulse frequency were set to 250 µs and 40 Hz, and the pulse intensity was set to the minimal value that can produce full extension and flexion motion. The only variable was the number of pulses, which was controlled according to the tremor amplitude. The experimental evaluation showed that the proposed system achieved 67% ± 13% suppression in tremor amplitude in six out of seven subjects. The one exception indicates that the out-of-phase strategy may not applicable in all cases.

Lastly, in a recent study, Dosen [78] investigated the feasibility of using sensory stimulation (below the motor threshold) to suppress tremor, and compared it with the stimulation above motor threshold. This experiment was tested on the wrist and finger flexors and extensors of six subjects with tremor. Five subjects presented tremor reduction in the range of 46–81% and 35–48% for the motor threshold and sensory threshold stimulation, respectively. In one patient, the tremor was not suppressed by the system. In all of the aforementioned studies, the stimulation was controlled above each subject’s motor threshold. However, this study discovered an interesting finding that tremor could also be suppressed without the contraction of the affected muscles. The underlying neurophysiological mechanism of tremor suppression using sensory stimulation may involve the modulation of afferent pathways (Fig.2.2).

Following the discovery of the potential contribution of afferent pathways to tremor management by Dosen et al. [78], Heo et al. [83] tested the sensory electrical stimulation on postural
tremor suppression in 18 subjects with ET. The electrical stimulation was delivered to the flexor carpi radialis, extensor carpi radialis, biceps brachii, and triceps brachii with a constant pulse frequency of 100 Hz and pulse width of 300 $\mu$s. The pulse intensity was controlled at a constant value below each subject’s motor threshold. The averaged tremor suppression levels during stimulation phase were 60% in the index finger MCP joint and 40% in the wrist. Interestingly, the authors discovered that the tremor amplitude was still suppressed five minutes after the stimulation. The average suppression levels of the post-stimulation phase were 67% in the index finger MCP joint and 45% in the wrist. A similar study conducted by Dideriksen et al. [84] validated a new stimulation strategy with a varying stimulation intensity that was controlled under the level of eliciting maximum H-reflex. A total of nine subjects with tremor participated in this study. The pulse frequency and pulse width were controlled at constant values of 100 Hz and 400 $\mu$s. The overall tremor suppression level achieved was 52%.

Lastly, Hao et al. [85] applied sensory electrical stimulation on the superficial radial nerves of eight subjects with resting tremor. The pulse frequency and pulse width were set to 250 Hz and 200 $\mu$s. The radiating threshold of each subject was first measured as the stimulation base value for tremor suppression. The final pulse intensity for tremor suppression was selected between 1.5 and 1.75 times the radiating threshold. The result of the experiment showed an overall 61.56%
suppression ratio with the proposed method.

### 2.4.3 Vibration-based Tremor Suppression Devices

The last tremor management approach is the mechanical vibration system. Leblanc et al. [86] and Lavu et al. [87] proposed two active tremor suppression devices that use mechanical vibrators. The proposed vibration device was controlled to vibrate at a fixed frequency to counterbalance the tremor generated by a shaker. Although, the results of both studies indicated that the use of vibration can reduce the magnitude of the simulated tremor, the vibration suppression systems of both studies were fixed to the ground. Such configuration may not achieve the same level of performance when the vibrator was worn by human. Therefore, the use of mechanical vibration in a wearable tremor suppression device still requires further validation.

To address the aforementioned issue, Kazi et al. [7] developed a wearable smart vibration glove that incorporates a piezoelectric actuator to generate vibrations at the hand. This glove was tested on both a human hand and a benchtop tremor simulator with three different vibration frequencies. The result of this feasibility study showed that the vibration glove has a better tremor suppression performance on a human hand than on a benchtop simulator. However, since this study conducted the test on only one subject, such low sample size does not yield a convincing conclusion.

Lastly, a recent study led by Lora-Millan et al. [88] tested the performance of a piezoelectric vibration device on suppressing tremor in the upper limbs of 18 subjects with ET. The experimental protocol consisted of a number of stimuli with frequencies ranging from 50 Hz to 450 Hz. A total of seven piezoelectric actuators were placed over the fingers, the back of the hand, below the wrist and below the elbow. The results showed that mechanical vibration caused an increase in tremor amplitude in 9 to 13 of the subjects depending on the stimulating strategy, while only one to four subjects’ tremor was suppressed. The authors concluded that mechanical vibration does not systematically reduce the tremor in ET. Although, ET does not seem to respond to mechanical vibration, the question of whether parkinsonian tremor reacts to mechanical vibration still requires further investigation.
2.5 Tremor Simulation Devices for Validation of a WTSD

To validate the design of a WTSD, the first step is often to test in a simulation environment. A tremor simulator should be able to generate real tremor characteristics, such as force and motion. Although, the validation on a patient population would be ideal, a mechatronic tremor simulator would serve as a risk-free tool to identify any possible drawbacks in the design. In addition, it is more accessible and does not require research ethics board approval.

Within the field of mechatronic tremor simulators, there have been relatively few published papers on upper limb tremor simulation. Existing physical tremor simulation devices can be categorized into benchtop tremor simulators [19, 40, 89], which often reproduce tremor motion and force on a simplified physical human joint model, and wearable simulators [90, 91], which instead produce tremor in a healthy volunteer’s joints using external vibration system.

For the purpose of validating a WTSD, although a wearable simulator allows the researcher to test their WTSD directly on a human, the physical human–machine interface often places a challenge on the physical placement of the WTSD due to interference from the simulator. Moreover, ethics approval is always required, which often prolongs the iterative process of the WTSD development.

2.6 Discussion

In this chapter, a review of the state of the art in characterization of parkinsonian tremor, tremor estimation techniques and tremor suppression devices has been presented.

It was found that the tremor in the elbow, wrist and MCP joints in the direction of flexion–extension has been characterized; however, the characteristics of tremor in the proximal interphalangeal (PIP) joints and distal interphalangeal (DIP) joints are still to be determined. Nonetheless, tremor in the all directions should be characterized if the purpose is to fully suppress tremor motion.

In the review of the tremor estimation techniques, most of the proposed tremor estimators cannot be used for real-time control because none of them can predict tremor motion ahead of time. Therefore, the computation time of the control system, and the electromechanical delay of
2.6 Discussion

the actuator, inevitably compromise the suppression performance. To solve this issue, a real-time control that incorporates tremor prediction algorithm should be investigated.

Lastly, the review of the tremor suppression devices has shown the effort of the researchers in investigating different techniques to suppress tremor. Although all three techniques reviewed have shown promising results in suppressing tremor, they all have drawbacks that keep them from being used by the end users, such as bulkiness, weight, discomfort by stimulation, vibration numbness, etc. It was also identified that none of the devices that have been proposed for tremor suppression combined multiple suppression techniques. Hypothetically, with the development of a smart control system that effectively controls multiple suppression techniques, a WTSD that incorporates multiple suppression techniques could compensate for the limitations of each technique and provide the user with the benefits of the various modalities. The following chapter will introduce the characterization of parkinsonian hand tremor and a novel algorithm that separates the tremor signal and voluntary motion signal.
Bibliography


BIBLIOGRAPHY


Chapter 3

Tremor Characterization and Estimation


3.1 Abstract

Recent progress in wearable technology has made wearable tremor suppression devices (WTSDs) for Parkinson’s patients a potentially viable alternative solution for tremor management. So far, in contrast to wrist and elbow tremor, finger tremors have not been studied in depth despite the huge impact that they have on a patient’s daily life. In addition, more evidence has been found showing that the performance of current tremor estimators may be limited by their model order due to the multiple harmonics present in tremor. The aim of this paper is to characterize finger and wrist tremor in both the time and frequency domains, and to propose a high-order tremor estimation algorithm. Tremor magnitudes are reported in the forms of linear acceleration, angular velocity, and angular displacement. The activation of forearm flexor and extensor muscles is also investigated. The frequency analysis shows that parkinsonian tremors produce oscillations of the hand with pronounced harmonics. At last, a high-order weighted-frequency Fourier linear combiner-
based Kalman filter (WFLC-KF) is proposed. The percentage estimation accuracy achieved from the proposed estimator is \(96.3\% \pm 1.7\%\), showing average improvements of 28.5\% and 48.9\% over its lower-order counterpart and the WFLC. The proposed estimator shows promise for use in a WTSD.

### 3.2 Introduction

Pathological tremor can be defined as involuntary and oscillatory movements. Tremor is commonly more severe in distal joints (\textit{i.e.}, fingers) than proximal joints (\textit{i.e.}, shoulders) [1, 2]. In general, tremor can be categorized into physiological tremor and pathological tremor. Pathological tremor can be further divided into resting tremor, such as parkinsonian tremor, and action tremor, such as essential tremor. Physiological tremor is a benign inherent component of voluntary movement. Its magnitude is negligible compared to the magnitude of voluntary movement. The suppression of physiological tremor is commonly seen in tasks that require precise movement, such as those performed during microsurgery [3]. Pathological tremor, in contrast, significantly affects the daily life of individuals, creating difficulties for fine motor tasks, including eating and writing. In addition, pathological tremor may cause social embarrassment leading to the avoidance of social activities.

Resting tremor occurs when no muscle contractions are made, while action tremor occurs during muscle contraction. Action tremor can then be further divided into postural tremor [4], which occurs when holding a specific posture against gravity, and kinetic, which occurs during motion. The management of these pathological tremors has varying degrees of effectiveness depending on the chosen approach. Medication is the mainstay of treatment; however, it often causes side effects and shows decreased effectiveness over time [5, 6]. In cases of severe tremor, surgical interventions, such as deep brain stimulator implantation and stereotactic thalamotomy [7, 8], have been performed. While these treatments are often very effective, they carry a risk of significant complications for the individual. These drawbacks have encouraged the development of an alternative approach for tremor management, such as a wearable tremor suppression device (WTSD). Unlike traditional treatments, a WTSD suppresses tremor according to the electrical and mechanical features of the tremor. Position, velocity, acceleration, force and electromyography (EMG) signals
are the most common forms of biofeedback that can be used as input to a WTSD.

Since Rosen et al. [9] first demonstrated that pathological tremor can be suppressed selectively, several research groups have proposed WTSDs as a potentially viable alternative solution for tremor management [10–25]. There are two main approaches used for WTSDs: one that applies mechanical loads [10–15, 24, 25], and another that applies functional electrical stimulation [16–23]. Both methods have shown promising results for suppressing tremor, although size, weight, power, and suppression magnitude and accuracy are still challenges for the clinical use of WTSD.

In the development of a WTSD, the characterization of the tremor motion and voluntary motion determines its design requirements. Several studies [26–35] have been conducted to characterize the mechanical and electrical features of tremor. Most of these studies have focused on elbow and wrist tremor, with very few studies characterizing the tremor found in the fingers, specifically the metacarpophalangeal (MCP) joint. The reported tremor frequencies from the elbow and wrist lie in the range of 3–12 Hz with the tremorous motion approximated by a sinusoidal motion. Tremor frequency tends to stay constant with slight variations; however, the amplitude of the tremor varies from person to person [1, 26, 32, 33]. As for the voluntary motion at hand, the reported frequency of activities of daily living is below 3 Hz [36].

In order to suppress tremor, it is important to extract the tremor movement from a combined movement, i.e., one that combines voluntary and tremor motion. Filters are commonly used in this process. A few studies have been conducted on the use of signal filtering for tremor estimation [23, 37, 38]; however, the time delay introduced by the filter impedes tremor control because an actuator cannot oppose the estimated tremor with zero-phase delay. This drawback of the common filter prevents its use for real time tremor suppression.

As an alternative, adaptive filtering is well suited for tremor estimation, as it is constantly adapting its parameters to the input signal based on a learning algorithm [39]. A Fourier linear combiner (FLC) was proposed by Vaz et al. [40, 41] based on the assumption that tremor can be simplified to a periodic signal. This suggests that tremor can be modeled by a sinusoidal or Fourier series. The FLC estimates tremor based on a known frequency. The least mean squares algorithm was incorporated to update the parameters, without involving a large computational workload [40], and it was able to surpass any common filter with its zero-phase feature [41].
In order to effectively suppress tremor, both the frequency and amplitude must adapt to changes. A Weighted-frequency Fourier Linear Combiner (WFLC) has been proposed to address this issue [42]. The computational load of this algorithm is low for real-time tremor suppression due to its simple form, i.e., no differential equations are involved. It adaptively estimates the frequency and amplitude of tremor by changing the filter weights according to a gradient search method (67% tremor reduction rate). It also assumes that the tremor signal can be modeled as a periodic signal, i.e., as a sine wave. In practice, a WFLC is often cascaded with a FLC, while some studies have also shown the feasibility of using a WFLC alone [10, 43, 44]. The main drawback of the WFLC is that it can only track a single dominant frequency. The performance degrades if multiple frequencies exist in the input signal.

The WFLC estimates tremor based on the estimation of the gradient of the mean square error with an assumption that the tremor is a periodic signal. However, tremor is not periodic. In contrast, a Kalman filter (KF) is expected to be more robust for nonperiodic motions, since it does not require any a priori assumption and it computes the optimal solution by minimizing the covariance of the a posteriori estimation error [43]. Recent studies have also shown that the adoption of a KF achieved better accuracy in tremor estimation than the gradient search method [43–45].

In a previous study [44], our group proved that the performance of a high-order WFLC-based KF exceeds its lower-order counterpart and that of the WFLC; however, the proposed architecture was complex and the computational load was excessive. The focus of this paper includes characterization of parkinsonian hand tremor and improvement of the design of the high-order WFLC-based KF (HWFLC-KF) for tremor estimation. The performance of the proposed estimator is evaluated and compared with the WFLC and its lower-order counterpart using real parkinsonian tremor data. The results show an improvement in tremor estimation over the WFLC ($p < 0.0001$), the lower-order counterpart ($p < 0.0001$) and the estimator presented in [44] ($p < 0.0001$).
3.3 Methods

Since many of the basic daily activities that involve hand motion can be achieved by using only the index finger, thumb and wrist, tremors from the thumb metacarpophalangeal (MCP) joint, index finger MCP joint and the wrist were first collected under experimental conditions from individuals with Parkinson’s disease (PD) using digital inertial measurement units (IMUs) and an electromagnetic tracking device. In addition to parkinsonian tremor, essential tremor is another important tremor to address. Due to an established collaboration with the Movement Disorders Clinic at Western University, the natural focus of this work was on patients with PD. Subjects with essential tremor may be recruited in a future study. The collected signals were then analyzed in both the time and frequency domains. Lastly, a tremor estimator was designed and validated using the collected tremor data. In this section, the experimental procedure, data processing and a high-order WFLC-based tremor estimation scheme are presented.

3.3.1 Subjects and Experimental Procedure

Approval for this study was obtained from the University of Western Ontario’s Human Research Ethics Board prior to the start of the trials. A total of 18 subjects with PD participated in the study. All subjects were diagnosed and recruited by a movement disorders neurologist. There were 11 males and 7 females. Their ages range from 60 to 84 (mean ± standard deviation is 69 ± 7) and 16 out of 18 patients were on medication (Levodopa, Pramipexole, and Amantadine). Most of the subjects were taking medication to address their symptoms. The medication was helpful for improving the bradykinesia and rigidity, but was not effective or poorly effective in controlling the tremor, which is not uncommon in tremor-dominant Parkinson’s disease. Prior to the experiment, each participant was given an explanation of the study, including the objectives and details of the study, data acquisition instruments, risks, benefits and confidentiality of the participant’s data. After the participant provided consent, the experiment started with an evaluation of the severity of their Parkinson’s disease. The neurologist administered the MDS-Unified Parkinson’s Disease Rating Scale (MDS-UPDRS): Part III (Motor Assessment) (1) to quantify the severity of the disease [46], including the severity of the tremor, and to select the hand with the greatest amount
of tremor for data collection. The MDS-UPDRS is a standard score for evaluating Parkinson’s disease. It evaluates all aspects of PD and rates each symptom on a 0 to 4 scale. The tremor is rated for severity (i.e., amplitude) of the tremor at rest (hands resting on the lap, legs at rest, and jaw tremor), posture (hands held out stretched) and action (hands moving from the finger to nose). The resting tremor component is also rated for constancy. All of the subjects presented tremor typical of Parkinson’s disease.

For tremor severity ratings, ordinal values of \{0, 1, 2, 3, 4\} were given, corresponding to \{no tremor, slight, mild, moderate, severe\}. Tremor severity across the 18 subjects was \(2.72 \pm 0.67\) and \(1.56 \pm 0.92\) for resting tremor and postural tremor, respectively. The total rating value of Items 3.15 to 3.18 from the MDS-UPDRS is \(8.72 \pm 2.47\). Seventeen right hands and one left hand were used for data collection based on tremor severity.

After the evaluation, a motion-sensing device was affixed over the patient’s hand (Fig. 3.1). This device includes five 9-axis IMUs (STEVAL-MKI108V2, STMicroelectronics®) and four 5 degree-of-freedom (DOF) electromagnetic (EM) motion trackers (Aurora, NDI®). The motion of each joint was acquired differentially by the two sensors on each side of the joint. The motions of the index finger MCP joint and thumb MCP joints were measured using both the IMU and the EM trackers. The wrist motion was measured only with the IMU. The distance between the accelerometer and the joint of interest was set to 5.5 cm for all patients. The EM field generator was placed in front of the subject’s hand to provide a low-intensity, varying EM field for the EM trackers. Lastly, an eight-channel EMG data collection armband (Myo, Thalmic Labs Inc.) was placed on the skin surface of the subject’s forearm muscles. The total weight of the wearable sensing device on the hand was 38 g. The impact of the weight loading on each subject’s hand tremor was considered to be negligible in this study.

In order to measure the resting tremor and postural tremor from the joints of interest, and to limit the impact of the movements from other body parts on the hand, the subject’s forearm was securely strapped to the experimental table, with the wrist hanging from the edge of the table. This setup allows only the hand to move freely in space. During the trial, each subject was asked to perform three tasks with their hands, as follows:

1. Keep the hand at rest on the table in a palm up position, Fig. 3.2(a).
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Figure 3.1: Experimental setup. The sensor placement is shown on a subject’s right hand. The Aurora field generator is placed in front. Due to the capability of the microcontroller, each microcontroller can record data from up to three IMUs.

2. Keep the hand at rest in a palm down position, Fig. 3.2(b).

3. Hold the hand up at an angle of approximately 45° to the table surface, Fig. 3.2(c).

During each task, data were recorded for about 60 seconds. While the subject was performing these tasks, he/she was asked simple questions (e.g., simple math questions) to distract their attention from suppressing their hand tremor. During all of the tasks, the subject was seated on a chair while the data were recorded. The entire experimental procedure was also video recorded. No patient identifiers were collected as part of this study. Considering that it was desired to distinguish the effect of the type of tremor, as well as the characteristics of the tremor at the different joints, the experiment was designed as a full factorial $2 \times 3$ repeated measures study with linear acceleration and angular velocity as the outcomes, and as a $2 \times 2$ repeated measures study with angular displacement as the outcome.
3.3 Methods

Figure 3.2: The three different hand positions used in the trials: (a) hand at rest in palm up position; (b) hand at rest in palm down position; (c) hand at 45° above the table surface.

3.3.2 Data Recording and Processing

All data were collected directly onto a PC (ASUS® desktop PC M51AC Series, Windows 8) using customized software. This software was developed in C++ using Visual Studio 2013. It was designed to manage data transfer and storage from all of the sensors to the PC. The software also ensured that all of the collected data were synchronized to the same timestamp. The IMUs were interfaced with an STC89C52RC microcontroller through the I2C protocol at 100 Hz. The data were then sent to the PC through a serial to USB adapter. Each IMU incorporates one 16-bit 3-DOF accelerometer, one 16-bit 3-DOF gyroscope and one 16-bit 3-DOF magnetometer. The data from the Aurora system were recorded using the Aurora application program interface (API) at its maximum sampling frequency of 40 Hz. The EMG data from the Myo system were sampled at 200 Hz and sent directly to the PC through a wireless Bluetooth connection.

Data preprocessing and analysis were conducted using MATLAB software (Version R2013a, The Mathworks, Inc.). The data preprocessing includes calibration of the accelerometers, which needed to occur prior to the start of each trial. The calibration equations are as follows:
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\[ CM = \left( A_r^T \cdot A_r \right)^{-1} \cdot A_r^T \cdot A_g \]

\[ CM = \begin{bmatrix} CM_1 \\ \cdots \\ CM_2 \end{bmatrix} = \begin{bmatrix} m_{11} & m_{21} & m_{31} \\ m_{12} & m_{22} & m_{32} \\ m_{13} & m_{23} & m_{33} \\ \cdots & \cdots & \cdots \\ m_{10} & m_{20} & m_{30} \end{bmatrix}, \]

\[ A_c = CM_1^T \cdot A + CM_2^T \]

where \( CM \) is the calibration matrix, \( A_g \) is the known Earth gravity matrix, \( A_r \) is the sampled data in line with the known Earth gravity matrix, \( A_c \) is the calibrated data, and \( A \) is the raw signal. This calibration process reduced the Root Mean Square (RMS) error from 0.33 m/s² to 0.007 m/s².

Following the calibration process, the tremor signal was extracted using a gravity compensation technique [47]. The algorithm is given below:

\[ g(i) = a \times g(i - 1) + (1 - a) \times A_c(i), g(0) = \begin{bmatrix} 0 \\ 0 \\ 0 \end{bmatrix} \]

\[ A_L(i) = A_c(i) - g(i), i \in \begin{bmatrix} 1 \\ n \end{bmatrix} \]

where \( g(i) \) is the filtered gravity component, \( a \) is the filter convergence coefficient, \( i \) is the current time step, \( n \) is the end of a signal with finite length, \( A_L \) is the linear acceleration caused directly by the tremor motion. Following the gravity compensation, the following equations were used to calculate the tremor magnitude.

\[ S_t(i) = \sqrt{S_x^2(i) + S_y^2(i) + S_z^2(i)} \]

\[ S_{rms} = \sqrt{S_t^2(1) + S_t^2(2) + \cdots + S_t^2(n)} \]

where \( S_x \), \( S_y \) and \( S_z \) are the amplitudes of the tremor motion from all three axes, \( S_t(i) \) is the tremor
3.3 Methods

magnitude, and \( S_{\text{rms}} \) is the RMS value of the tremor magnitude. With the linear acceleration obtained, the joint tremor motion was calculated using the data from the two sensors located on each side of the joints.

3.3.3 Enhanced High-Order WFLC-based Kalman Filter

The characterization of tremor and the investigation of the treatment response can be performed using a recorded data set without requiring the input of new data. Such analysis requires quantification tools to process the data collected from sensors [48]. In contrast, tremor estimation must be performed while new input data are being acquired. It offers the advantage of estimating tremor motion in real time, thereby allowing tremor to be actively controlled by an external assistive orthosis. In order to precisely estimate tremor motion, the estimator must be able to eliminate noise and voluntary motion. In this section, an enhanced version of an HWFLC-KF is proposed based on the previous work [44]. In the previous version, the first stage of the estimator consisted of multiple band-pass filter (BPF) to break a tremor signal into multiple frequency bands. The frequency of each tremor harmonic was then estimated by a WFLC. Lastly, a KF was used to estimate and combine each estimated harmonic to form the estimated tremor. The enhanced version presented herein simplifies the complex architecture of the previous version by introducing a frequency adaptation gain vector. This vector allows the algorithm to estimate the frequency of each harmonic separately without using multiple band-pass filters and WFLCs. In addition, this algorithm can be easily applied to estimate signals with \( n \) harmonics.

Recent studies [49–51] on parkinsonian tremor have suggested that it consists of multiple harmonics. The frequency of the \( k \)th harmonic is roughly \( k \) times the fundamental frequency. However, most of the tremor estimators proposed in the past use a model that tracks only the single dominant frequency [28, 42, 52]. This is insufficient to achieve good estimation performance. In order to emphasize the importance of the higher harmonics in parkinsonian tremor and increase the estimation accuracy, an enhanced high-order WFLC-based (HWFLC) KF is introduced.

The architecture of the enhanced HWFLC is similar to the WFLC. The state vector \( \mathbf{X}(i) = \begin{bmatrix} x_1(i), \ldots, x_{2M}(i) \end{bmatrix}^T \) contains the time varying frequency \( (\omega_j) \) of the estimated signal (Eq. 3.4), which is computed using the least mean squares (LMS) algorithm,
3.3 Methods

\[ x_j(i) = \begin{cases} 
\sin \left( \sum_{t=0}^i \omega_j(t) \right), & 1 \leq j \leq M \\
\cos \left( \sum_{t=0}^i \omega_{j-M}(t) \right), & M + 1 \leq j \leq 2M 
\end{cases} \tag{3.4} \]

where \( M \) is the number of harmonics that constitute the model, and \( i \) is the current time step. At each iteration, \( \omega_j \) and the amplitude weight \( P(i) = \begin{bmatrix} p_1(i) & \cdots & p_{2M}(i) \end{bmatrix}^T \) are calculated using Eqs. 3.5 - 3.7,

\[ \epsilon(i) = S_t(i) - \sum_{j=1}^M \left[ p_j(i) \cdot \sin \left( \sum_{t=0}^i \omega_j(t) \right) + p_{j+M}(i) \cdot \cos \left( \sum_{t=0}^i \omega_j(t) \right) \right] \tag{3.5} \]

\[ \omega_j(i+1) = \omega_j(i) + 2 \cdot v_j \cdot \epsilon(i) \cdot [p_j(i) \cdot x_{M+j}(i) - p_{M+j}(i) \cdot x_j(i)], \quad j \in [1, M] \tag{3.6} \]

\[ P(i+1) = P(i) + 2 \cdot \mu \cdot X(i) \cdot \epsilon(i) \tag{3.7} \]

where \( \epsilon(k) \) is the estimation error, \( S_t(k) \) is the band-pass filtered signal, and \( \mu = \text{diag} \{ \mu_1 \cdots \mu_{2M} \} \) and \( v = \{ v_1 \cdots v_{2M} \} \) are the amplitude and frequency adaptation gains, respectively. The reference model used in the WFLC considered tremor as a combination of a series of harmonics, with their frequencies in an arithmetic progression; however, the frequencies of the tremor harmonics do not meet this requirement. Other WFLC derivatives [53–55] use multiple frequency bands to increase the estimation accuracy. Although the results are acceptable, the computation is rather expensive. In comparison, the enhanced HWFLC estimates each harmonic separately with varying frequency and the computational load is fairly low.

In the enhanced HWFLC-KF, the HWFLC estimates the frequencies of the tremor with the advantage of the WFLC’s frequency tracking. The KF uses the estimated harmonic frequencies from the enhanced HWFLC and the band-pass filtered signal to construct the estimated tremor. Suppose that the tremor signal can be modeled as a third-order harmonic model \((S_t(i))\), as in Eq. 3.8.
3.3 Methods

\[ S(t) = S_{t1}(i) + S_{t2}(i) + S_{t3}(i) \]

\[ S_{t1}(i) = a_{n1}(i) \cdot \sin(2\pi f_{t1} iT_s) + b_{n1}(i) \cdot \cos(2\pi f_{t1} iT_s) \]

\[ S_{t2}(i) = a_{n2}(i) \cdot \sin(2\pi f_{t2} iT_s) + b_{n2}(i) \cdot \cos(2\pi f_{t2} iT_s) \]  \hspace{1cm} (3.8)

\[ S_{t3}(i) = a_{n3}(i) \cdot \sin(2\pi f_{t3} iT_s) + b_{n3}(i) \cdot \cos(2\pi f_{t3} iT_s) \]

where \( S_{t1}, S_{t2}, \) and \( S_{t3} \) are the first, second, and third harmonics, \( a_{n1}, a_{n2}, a_{n3}, b_{n1}, b_{n2}, \) and \( b_{n3} \) are the Fourier coefficients, \( f_{t1}, f_{t2}, \) and \( f_{t3} \) are the estimated frequencies of the corresponding harmonics from the enhanced HWFLC, and \( T_s \) is the time step. Based on the model, the state transition matrix \( (F(i)) \) of the KF can be defined as Eq. 3.9.

\[
\begin{bmatrix}
S_{t1}(i+1) \\
\dot{S}_{t1}(i+1) \\
\vdots \\
S_{tM}(i+1) \\
\dot{S}_{tM}(i+1)
\end{bmatrix}_{1\times2M} = F(i) \cdot 
\begin{bmatrix}
S_{t1}(i) \\
\dot{S}_{t1}(i) \\
\vdots \\
S_{tM}(i) \\
\dot{S}_{tM}(i)
\end{bmatrix}_{1\times2M}
\]

(3.9)

\[
F_j(i) = 
\begin{bmatrix}
\cos(2\pi f_j(i) T_s) & \sin(2\pi f_j(i) T_s) / (2\pi f_j(i) T_s) \\
-2\pi f_j(i) T_s \sin(2\pi f_j(i) T_s) & \cos(2\pi f_j(i) T_s)
\end{bmatrix}, j \in [1, M]
\]

The tuned KF parameters are given in Table 3.1. The architecture of the enhanced HWFLC-KF is presented in Fig. 3.3. The value of \( M \) was set to 3, in order to estimate the fundamental, 2\textsuperscript{nd} and 3\textsuperscript{rd} harmonics. Unlike the WFLC, the frequencies of the tremor’s harmonics are estimated separately. Therefore, the enhanced HWFLC is well suited to providing accurate tremor frequency components to the KF for tremor estimation.
Table 3.1: Parameters of the HWFLC-KF.

<table>
<thead>
<tr>
<th>Par.</th>
<th>Value</th>
<th>Par.</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\nu$</td>
<td>1e-5, 1e-5, 1e-7</td>
<td>$\mu$</td>
<td>diag ${0.16, 0.05, 0.0165}$</td>
</tr>
<tr>
<td>$R$</td>
<td>diag ${400, 800, 1600}$</td>
<td>$T$</td>
<td>0.01 s</td>
</tr>
<tr>
<td>$Q$</td>
<td>diag ${0.9, 0.9, 2, 2, 3, 3}$</td>
<td>$H$</td>
<td>$\begin{bmatrix} 1 &amp; 0 &amp; 0 &amp; 0 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 1 &amp; 0 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 0 &amp; 0 &amp; 1 &amp; 0 \end{bmatrix}$</td>
</tr>
<tr>
<td>$G$</td>
<td>$\begin{bmatrix} 10^{-4} &amp; 10^{-7} &amp; 0 &amp; 0 &amp; 0 &amp; 0 \ 10^{-7} &amp; 10^{-4} &amp; 0 &amp; 0 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 10^4 &amp; 10^1 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 10^1 &amp; 10^4 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 0 &amp; 0 &amp; 10^{-5} &amp; 10^{-8} \ 0 &amp; 0 &amp; 0 &amp; 0 &amp; 10^{-8} &amp; 10^{-5} \end{bmatrix}$</td>
<td>$\begin{bmatrix} 10^{-5} &amp; 10^{-8} \ 10^{-8} &amp; 10^{-5} \end{bmatrix}$</td>
<td></td>
</tr>
</tbody>
</table>

To evaluate the performance of the enhanced HWFLC-KF, the raw tremor data collected from the trial described in Section 3.3.1 were used. A Fast Fourier Transform (FFT) of all raw data was performed to identify the cutoff frequency of filtering out voluntary motion, the result showed that 1 Hz cutoff frequency is sufficient to eliminate the voluntary motion. Prior to the evaluation, a 2$^{nd}$-order Butterworth zero-phase band-pass filter [26] with cutoff frequencies of 1 and 30 Hz was used to extract the tremor signal. Lastly, the Percentage Estimation Accuracy (PEA) and the correlation coefficient were used to compare the performance, as Eq. 3.10.

$$\text{PEA} = \left(1 - \frac{\| \text{Estimated value} - \text{True value} \|}{\text{True value}} \right) \times 100\% \quad (3.10)$$

To compare the computational performance of the proposed estimator, the percentage computation time difference was calculated using Eq. 3.11,

$$\text{Pct. Comput. Time Diff.} = \frac{T_p - T_o}{T_o} \times 100\%, \quad (3.11)$$

where $T_p$ represents the elapsed CPU time of the proposed estimator required to process one patient data set and $T_o$ represents the elapsed CPU time required by the compared estimator to process one patient data set.
3.3 Methods

Figure 3.3: Signal flow diagram of the HWFLC-KF. It consists of a modified WFLC (HWFLC) and a KF. The HWFLC estimates the frequency of each tremor harmonic separately. The estimated frequencies are then used in the KF to generate the state transition matrix.

3.3.4 Statistical Analysis

The results were first analyzed using a histogram to determine the data distribution. Since the data appeared to be normally distributed, a repeated measures comparison was performed to determine statistical significance. A Bonferroni adjustment was applied to the comparison. The IBM Statistical Package for the Social Sciences (SPSS, v.24) was used to complete the analysis. A p value of less than 0.05 was considered to be statistically significant.
3.4 Results and Discussion

3.4.1 Parkinsonian Tremor Characterization

This section presents the collected parkinsonian hand tremor data, which were analyzed in both the time and frequency domains. The index finger, thumb and wrist tremor were collected and characterized in terms of linear acceleration, angular velocity and angular displacement. In addition, the activation of the forearm flexor and extensor muscles was also investigated.

Fig. 3.4 shows a typical resting tremor signal (left side) and a postural tremor signal (right side) in the time domain. The tremors from the index finger (IF) MCP joint, the thumb (T) MCP joint and the wrist (W) joint are shown in the blue, red and purple curves, respectively. It is important to point out that the tremorous oscillations come as clusters with different periods. An example is shown in the bottom figures in Fig. 3.4. The signals with black lines underneath mark the tremor clusters. Each cluster is identified with an onset and a drop-down. The onset of a cluster (\(t_0\) and \(t_2\)) is often observed as an increase in the amplitude of the non-voluntary motion. The increased amplitude persists until it is reduced by a large degree, which indicates the drop-down state (\(t_1\) and \(t_3\)). After this cluster, tremor from the affected body segment presents a pseudo-tremor-free status (as indicated by the green lines underneath the signals in Fig. 3.4) until the next cluster occurs.

The quantification of tremor starts with the identification of tremor clusters. A 1-second long window was used to sweep through the entire signal. The data from the cluster(s) were used to calculate the tremor RMS magnitude. The linear acceleration and angular velocity acquired from the IMUs, and the angular displacements acquired from the Aurora system are presented in Table 3.2. All 18 subjects presented resting tremor, while 13 subjects presented postural tremor (the other five subjects’ data were excluded from the quantification analysis of postural tremor due to the absence of tremor in each joint). The RMS angular displacement of the wrist tremor is not presented in this study, as the number of EM trackers was limited to four.

The RMS magnitudes (linear acceleration and angular velocity) of both resting tremor and postural tremor show that tremor magnitude diminishes from the distal joints to the proximal joints. The tremor at the wrist presented a lower linear acceleration and angular velocity compared
3.4 Results and Discussion

Figure 3.4: Time series of a sampled tremor signal. The left side shows the resting tremor, the right side shows the postural tremor. The tremor in the index finger is shown in blue, the thumb is shown in red, and the wrist is shown in purple. In the bottom figure, the tremor signal segments with black lines underneath represent tremor clusters, which are separated by pseudo-tremor-free status (indicated by green lines underneath). The sampled tremor clusters are shown over a 30 s time window. $t_0$, $t_1$, $t_2$ and $t_3$ represent the time of tremor onset, the end of the 1st tremor cluster, the beginning of the 2nd tremor cluster, and the end of the 2nd tremor cluster, respectively.

Table 3.2: RMS magnitude of parkinsonian tremor.

<table>
<thead>
<tr>
<th></th>
<th>Resting Tremor (Mean ± SD) n = 18</th>
<th>Postural Tremor (Mean ± SD) n=13</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>RMS Linear Acceleration (m/s²)</strong></td>
<td>IF MCP Joint: 2.1 ± 1.2</td>
<td>IF MCP Joint: 2.6 ± 1.0</td>
</tr>
<tr>
<td></td>
<td>T MCP Joint: 2.1 ± 1.4</td>
<td>T MCP Joint: 3.0 ± 1.6</td>
</tr>
<tr>
<td></td>
<td>Wrist: 0.9 ± 0.5</td>
<td>Wrist: 1.5 ± 0.7</td>
</tr>
<tr>
<td><strong>RMS Angular Velocity (°/s)</strong></td>
<td>IF MCP Joint: 58.5 ± 34</td>
<td>IF MCP Joint: 77.7 ± 38.2</td>
</tr>
<tr>
<td></td>
<td>T MCP Joint: 49.9 ± 35.5</td>
<td>T MCP Joint: 64.0 ± 32.5</td>
</tr>
<tr>
<td></td>
<td>Wrist: 35.6 ± 20.9</td>
<td>Wrist: 48.8 ± 23.2</td>
</tr>
<tr>
<td><strong>RMS Angular Disp. (°)</strong></td>
<td>IF MCP Joint: 2.9 ± 1.8</td>
<td>IF MCP Joint: 3.1 ± 1.3</td>
</tr>
<tr>
<td></td>
<td>T MCP Joint: 2.2 ± 1.6</td>
<td>T MCP Joint: 2.4 ± 1.2</td>
</tr>
<tr>
<td></td>
<td>Wrist: —</td>
<td>Wrist: —</td>
</tr>
<tr>
<td><strong>AFFE Ratio</strong></td>
<td>1.21 ± 0.47</td>
<td>0.93 ± 0.68</td>
</tr>
</tbody>
</table>
3.4 Results and Discussion

to the index finger and thumb. The tremor from the thumb presents a lower mean angular displacement ($p = 0.007$) compared to the tremor from the index finger. The linear acceleration and the angular velocity from the index finger and the thumb were not significantly different. Following the spatial tremor comparison, the mean magnitudes of the postural tremor are found to be greater, which is aligned with the literature [26, 56]; however, the repeated measures comparison showed that there are no significant differences between resting tremor and postural for linear acceleration ($p = 0.599$), angular velocity ($p = 0.730$), or angular displacement ($p = 0.217$). Interestingly, the mean amplitude of the resting tremor was higher in the MDS-UPDRS rating than during the test trials. This may be caused by the effect of biomechanical uncertainties of the different postures on the measurement of tremor motions, i.e., hand position change. The standard position for rating the resting tremor is on the lap, as this is when the hand is most relaxed, whereas the experimental setup used during data collection had the hand resting on a table. In this position, although the tremor was still present with a consistent frequency, the amplitude of the tremor was lower than when resting on the lap. This finding is supported by a recent study showing that position change can reduce resting tremor amplitude [57]. In Table 3.2, the tremor magnitudes were found to spread out to a large extent around the mean value. This is possibly a result of the variability in the severity of parkinsonian tremor across subjects. In other words, a large heterogeneity may have existed in the subject group, which may have resulted in a large difference in the tremor severity. The sensor’s intrinsic uncertainty and the subjects’ suppression of their tremor may also cause the tremor magnitudes to vary.

Lastly, the ratio between the activation of the forearm flexors and extensors, i.e., the AFFE ratio, during the tremor-on state was calculated (bottom row of Table 3.2) to quantify the contributions of the forearm flexors and extensors to the generation of tremor. Due to the inevitable existence of voluntary motion when recording the postural tremor, the actual postural tremor EMG signals from the flexors and extensors were approximated as the difference of the Myo outputs at tremor-on and tremor-off states. The mean value of the AFFE ratio is found to be greater than one for resting tremor, and smaller than one for postural tremor. This indicates that, in comparison with the postural tremor, the flexors are activated more by tremor than the extensors when the hand is at rest (resting tremor). However, the data set does not have enough power to
3.4 Results and Discussion

Table 3.3: Spatial tremor comparison.

<table>
<thead>
<tr>
<th></th>
<th>IF MCP Joint</th>
<th>T MCP Joint</th>
<th>Wrist</th>
</tr>
</thead>
<tbody>
<tr>
<td>Linear Acceleration</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(m/s²)</td>
<td>—</td>
<td>1</td>
<td>&lt; 0.05</td>
</tr>
<tr>
<td></td>
<td>1</td>
<td>—</td>
<td>&lt; 0.05</td>
</tr>
<tr>
<td></td>
<td>&lt; 0.05</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Angular Velocity</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(°/s)</td>
<td>IF MCP Joint</td>
<td>0.067</td>
<td>0.022</td>
</tr>
<tr>
<td></td>
<td>T MCP Joint</td>
<td>—</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Wrist</td>
<td>0.0091</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>0.022</td>
<td>—</td>
</tr>
<tr>
<td>Angular Disp.</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(°)</td>
<td>IF MCP Joint</td>
<td>0.007</td>
<td>—</td>
</tr>
<tr>
<td></td>
<td>T MCP Joint</td>
<td>—</td>
<td>—</td>
</tr>
</tbody>
</table>

Table 3.4: Spatial tremor comparison.

<table>
<thead>
<tr>
<th></th>
<th>Resting Tremor</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Linear Acceleration</td>
</tr>
<tr>
<td>Postural Tremor</td>
<td></td>
</tr>
<tr>
<td>Linear Acceleration</td>
<td>0.599</td>
</tr>
<tr>
<td>Angular Velocity</td>
<td>—</td>
</tr>
<tr>
<td>Angular Disp.</td>
<td>—</td>
</tr>
</tbody>
</table>

show a significant difference ($p = 0.165$). Based on the t test with a selected power of 0.9 when the significance level is 0.05, the minimum sample size was calculated as 32. Therefore, a larger subject group is needed to prove this hypothesis. The complete pairwise comparisons are given in Table 3.3 and Table 3.4.

In the case of the tremor that showed only a single harmonic, the mean fundamental frequency was found to be higher than the mean fundamental frequency of the tremors with multiple harmonics. In contrast to the multiple harmonics tremor, the frequency of the single harmonic is increased due to the reduced number of harmonics, *i.e.*, concentrated power on the fundamental harmonic. The frequency of the postural tremor lies within the range of 5.3 Hz to 7.7 Hz, and the frequency of the resting tremor is around 5.2 Hz.

In most of the previous studies [24, 58], parkinsonian tremor was considered and reported as a mono-harmonic signal, its fundamental frequency lying within the range of 4 to 12 Hz. However, Fig. 3.5 shows that parkinsonian tremors produce oscillations with pronounced harmonics. The frequency of the $k^{th}$ harmonic is roughly $k$ times the fundamental frequency. This result is in line
with the results from other studies [51, 59]. The discovery of this pattern may facilitate the design of tremor suppression devices by improving the control algorithm.

An analysis of the power distribution was conducted on the tremor from each joint in order to evaluate the contribution of each harmonic. The percentage power of the 2\textsuperscript{nd} and 3\textsuperscript{rd} harmonics to the fundamental harmonic is presented in Fig. 3.6. This result was acquired from the 17 subjects that showed multiple harmonics in their resting tremor, and from the 11 subjects that showed multiple harmonics in their postural tremor. For the resting tremor, the mean percentage power of the 2\textsuperscript{nd} and 3\textsuperscript{rd} harmonics of all joints are 60.9\% and 38.9\%, respectively, and for the postural tremor are 50.3\% and 20.9\% for the 2\textsuperscript{nd} and 3\textsuperscript{rd} harmonics, respectively. These results show that the 2\textsuperscript{nd} and 3\textsuperscript{rd} harmonics are so strong that they cannot be excluded. In addition, 12 out of 17 subjects have at least one joint in which the power of the 2\textsuperscript{nd} harmonic is 50\% higher than the fundamental harmonic. Based on the results obtained above, it is concluded that an accurate
3.4 Results and Discussion

Figure 3.6: Tremor power distribution. Top: resting tremor, bottom: postural tremor. Each subject is represented by a different color from blue to red. The mean values are shown in black. The histogram is sorted by the magnitude of the wrist’s second harmonic for better presentation.

parkinsonian tremor model should contain multiple harmonics.

3.4.2 Evaluation of the Enhanced HWFLC-KF

To show the results of the enhanced HWFLC-KF, Fig. 3.7 plots the tracking performance of the WFLC and the enhanced HWFLC-KF on the tremor data from one subject selected at random. The tremor signal (red) was extracted using a 2nd order digital zero-phase bandpass filter with cutoff frequencies of 1 and 30 Hz. The green and blue curves represent the estimated tremor signal from the WFLC and enhanced HWFLC-KF, respectively. The model order of both the WFLC and the enhanced HWFLC-KF was set to 3. It can be observed that the enhanced HWFLC has better amplitude tracking performance than the WFLC, while the frequency tracking performance
3.4 Results and Discussion

Figure 3.7: Time series of the tremor and estimated tremors from the WFLC and the HWFLC-KF. The tremor signal was extracted using a 2nd order digital zero-phase bandpass filter with cutoff frequencies of 1 and 30 Hz. The inset shows the signals between 4.5 s and 5.5 s.

is similar (WFLC and enhanced HWFLC-KF have PEAs of 47.4 ± 8.4% and 96.3 ± 1.7%, p < 0.001). Similar results were obtained for the other 17 subjects. The estimation error of the WFLC may be the result of a limitation of the gradient-based estimation algorithm. Different from the WFLC, the KF achieves an optimal estimation by minimizing the covariance of the a posteriori estimation error. The combination of the enhanced HWFLC and the KF obtains the advantages of enhanced HWFLC in frequency tracking and KF in amplitude estimation. Therefore, the proposed enhanced HWFLC-KF achieves higher tremor estimation accuracy than other approaches reported in the literature. However, one potential limitation of the proposed method is the cutoff frequency of the filter for voluntary motion filtration. In reality, some voluntary movements could be produced with frequencies in the same range as tremor. For these cases, the current approaches will have lower estimation performance, as they cannot distinguish tremor based on frequency features. Therefore, other features or artificial intelligence-based estimators are required to address this problem.

The results from the tremor analysis in the frequency domain have shown that parkinsonian tremor consists of multiple harmonics. Although the frequency of the $k^{th}$ harmonic is roughly $k$
times the fundamental frequency, the variation in the frequency of each harmonic should not be ignored. However, the WFLC considers the frequency of $k^{th}$ harmonic to be exactly $k$ times the fundamental frequency, and in most of the studies [21, 30, 42, 43, 45, 60] the WFLC was used to estimate only the fundamental frequency. These are potential sources of error for their estimation results. In contrast to the WFLC, the enhanced HWFLC-KF estimates the frequency of each harmonic separately. This modification likely results in better tracking performance.

The frequency components of the estimated tremor from the WFLC and the enhanced HWFLC-KF are given in Fig. 3.8. The model order of both estimators was set to 3. The green line represents the frequency distribution of the estimated tremor from the WFLC, and the blue line represents the estimated tremor from the enhanced HWLFC-KF. The tremor recorded from the patient is shown in red. It can be seen from the graph that the deviation between the WFLC and the recorded tremor increases with higher harmonics. This is likely due to the fact that the WFLC cannot track higher harmonics separately and adaptively, i.e., the estimated frequency from the higher-order component of the WFLC is an integral multiple of the estimated frequency from the first-order component. Therefore, the estimation error from the WFLC, when processing tremor with multiple harmonics, could be significantly increased. These results again indicate that the enhanced HWFLC-KF may achieve better performance for tremor estimation.

The effect of the order of the enhanced HWFLC-KF was also investigated. Fig. 3.9 shows the estimation errors of the higher order ($M = 3$) and the lower order ($M = 1$). Since the lower order estimator does not eliminate the higher harmonics, its estimation error presents pronounced components of higher harmonics (Fig. 3.9, top, red curve). In contrast, the high-order estimator eliminates most of the higher harmonics (Fig. 3.9, top, blue curve). The frequency distribution of both estimated tremor and the recorded tremor are shown in Fig. 3.9, bottom. Both estimated tremors present a similar power distribution to the recorded tremor for the $1^{st}$ harmonic. However, the $2^{nd}$ and $3^{rd}$ harmonics present a significant difference, which indicates that the lower-order estimator failed to estimate the presence of the higher harmonics. The comparison of the two estimators shows that the high-order estimator produces better estimation results. The same analysis was extended to all 18 subjects to draw a more robust conclusion.

Fig. 3.10 presents the estimation accuracy and correlation of the two estimators. The mean
3.4 Results and Discussion

Figure 3.8: Power spectrums of the tremor (red), and the estimated tremor from the enhanced HWFLC (blue) and WFLC (green). The model order of the WFLC was set to three in this comparison.

Figure 3.9: Estimation error of the estimated tremors from the higher-order and lower-order estimators.
estimation accuracy of the high-order and low-order estimators are 90.1% and 67.8%, and the mean correlation coefficients are 0.99 and 0.95. The average increase is 22.3% for accuracy and 0.04 for the correlation. Both were found to be significantly different ($p < 0.001$). The estimation performance of the proposed estimator is comparable with the other tremor estimators in recent literature (past six years), with an estimation accuracy range from 84% to 98% [50, 55, 61–63].

This improvement in estimation performance is likely facilitated by the fact that most of the parkinsonian tremors have multiple harmonics. The previous power distribution analysis has shown that 17 out of 18 subjects presented multiple harmonics, and that the 2$^{nd}$ and 3$^{rd}$ harmonics, especially the 2$^{nd}$, made a large contribution to the tremor. This implies that the use of a high-order model in the tremor estimator can result in better performance. It is also important to point out that one out of 18 patients presented a single harmonic. In this case, the estimation accuracies from both estimators are similar. As expected, the additional order did not improve the estimation accuracy.

Lastly, the computation time of the proposed estimator was compared with the estimator from our previous work [44] and the WFLC, as shown in Fig. 3.11. The mean ± standard deviation (SD) of the computation time for the proposed estimator, our previous work and the WFLC are
3.5 Conclusions

This paper studied finger and wrist tremor in both the time domain and frequency domains from 18 subjects with PD. In addition, an enhanced HWFLC-KF was proposed and evaluated using tremor data from all 18 subjects presenting with parkinsonian tremor. The quantified characteristics of parkinsonian hand tremor were given in the forms of linear acceleration, angular velocity, angular displacement and the flexor/extensor activation ratio. The magnitudes of the mechanical features of the tremor were found to be similar between the index finger MCP joint and thumb MCP joint, but both of them are significantly higher than wrist tremor. A common multiple harmonics pattern was observed in parkinsonian hand tremor. The frequencies of the harmonics present

![Computation time comparison](image)

Figure 3.11: Computation time comparison. Each blue dashed line connects the computation time the estimators used to estimate one patient tremor data set. A total of 18 dashed lines are shown, which represents all 18 patient tremor data sets. The red error bar shows the mean and standard deviation of the computation time of each estimator.

1.15 ± 0.65 s, 1.34 ± 0.64 s, and 0.65 ± 0.38 s. The proposed estimator is 16% ± 8% faster than the estimator from our previous work and 79% ± 8% slower than the WFLC. Although the proposed estimator requires more computation time than the WFLC, the application of this estimator holds the potential to improve the performance of a WTSD.
as an arithmetic progression. Further analysis showed that the second harmonic makes a large contribution to the tremor, although not as large as the 1st harmonic, therefore, it should not be ignored. These results are especially important when considering an optimal design of the actuation and control systems of a WTSD.

Based on the findings of the frequency pattern analysis, a tremor estimator was proposed. The results showed that the proposed estimator has better tremor estimation performance than the WFLC, and that the higher-order version of the estimator showed the best performance. This proposed estimator shows promise for use in a WTSD.
Bibliography


Chapter 4

First Generation WTSG


4.1 Development and Validation of a Finger Tremor Simulator

4.1.1 Abstract

Tremor, being one of the most severe symptoms of Parkinson’s disease, has been considered as not only a medical problem but also an engineering problem. Increasingly, wearable technologies are being considered as a viable treatment option. In order to study and control tremor in the field of engineering, the first step often includes modeling and simulation, as access to patients is limited. With the successful realization of a finger tremor simulator, a wearable tremor suppression device could be validated prior to testing on humans. In this study, a tremor simulator was designed and validated with recorded patient tremor data. Two experimental assessments were conducted on the validation of tremor motion reproduction and tremor torque reproduction. The results showed that the proposed simulator has 9%, 82% and 141% error on reproduction of the power of the 1st, 2nd and 3rd harmonics of the tremor, and 11.89% mean error on motion reproduction. The tremor
torque measured at the index finger metacarpophalangeal joint is 0.02 ± 0.02 Nm, and the output torque from the tremor simulator is 0.03 ± 0.01 Nm. Further parameter adjustment of the control system is required to improve performance.

4.1.2 Introduction

Parkinsonian tremor is often seen as an involuntary, rhythmic, oscillatory movement with varying amplitude and relatively stable frequency [1–3]. Clinically, parkinsonian tremor can be further categorized into two types: resting and postural tremor [4]. The measured frequency of parkinsonian hand tremor is within the range of 3.5 Hz to 11.5 Hz [3, 5]. A recent study has shown that parkinsonian hand tremor consists of multiple harmonics, and that the 2nd and 3rd harmonics cannot be ignored [3].

The current treatment for Parkinson’s disease (PD) includes both medication and brain surgery [6, 7]. However, a large proportion of patients’ tremor (about 25%) is not responsive to medication [8]; although brain surgery is effective in reducing tremor, it is often associated with a series of complications [9]. Considering the existing downsides of current treatments, alternative tremor management approaches have been studied since Rosen’s experiment on suppressing tremor with mechanical loading [10]. Additional results [11–18] have been reported that indicate that active suppression devices have become a viable alternative solution for tremor management.

To validate the design of a wearable tremor suppression device, a simulator that can generate real tremor characteristics is required. Although a patient with parkinsonian tremor would be ideal, a mechatronic device that reproduces recorded tremor data would serve as a first step for risk-free validation and would be much more accessible.

In this study, a finger tremor simulator was designed and validated. The proposed simulator is actuated by a brushless DC motor. Two experimental assessments were conducted to evaluate the performance of tremor motion and tremor torque reproduction. The overall objective of this study was to develop a reliable tremor simulator that can be used to evaluate a wearable tremor suppression device in the future.
4.1 Development and Validation of a Finger Tremor Simulator

4.1.3 Methods

In order to develop the tremor simulator, access to real tremor data was needed. The method for collecting these data is described in the following section.

4.1.3.1 Parkinsonian Tremor Data Collection

A total number of 16 subjects with parkinsonian tremor were recruited to assess the tremor motion and torque in their metacarpophalangeal joints (MCP) in the flexion–extension direction. Based on the severity of their hand tremor, 13 right hands and 3 left hands were chosen for data collection. All subjects signed a consent form prior to the start of the experiment. The experimental protocol was approved by the University of Western Ontario’s Research Ethics Board.

For motion data collection, two 5 degree of freedom (DOF) electromagnetic (EM) motion trackers (Aurora, NDI®) were affixed over the patient’s index finger’s metacarpal and proximal phalanges. Each patient was asked to perform two sets of tasks:

1. Hand resting on a table
2. Voluntary finger reciprocating flexion-extension motion

All signals were recorded directly onto a personal computer (PC) through the Aurora’s customized software interface. The sampling frequency was set to the maximum of 40 Hz.

For torque data collection, the distal end of the subject’s index finger proximal phalange was attached to a torque sensor (TRS605-FSH02056, FUTEK®) while the arm and hand were strapped on a table. All signals were recorded directly onto a PC through a data acquisition card (USB-6002, NI®). The sampling frequency was configured to 1 kHz. These signals were used to quantify the tremor torque and to compare with the torque output of the tremor simulator, as described in the next section.

4.1.3.2 Experimental Setup for the Tremor Simulator

The design and evaluation setup of the proposed tremor simulator is shown in Fig. 4.1. The simulator consists of a brushless DC motor with planetary gearhead (#6, EC-max 16, reduction
4.1 Development and Validation of a Finger Tremor Simulator

Figure 4.1: The tremor simulator and experimental setups for the assessments of tremor motion simulation (left) and tremor torque simulation (right). 1. Aurora magnetic field generator, 2. Aurora EM motion tracker, 3. data acquisition card, 4. torque sensor, 5. motor controller, 6. brushless DC motor, 7. 3D printed finger.

ratio 29:1, Maxon Motor) and a 3D printed finger (#7). The printed finger and motor are supported by two support structures. The motor shaft is directly connected to one side of the printed finger, and the other side is supported by two ball bearings and a dowel pin.

In order to evaluate the performance of tremor motion simulation, an experimental setup was built as shown in the left side of Fig. 4.1. An EM motion sensor (#2) was attached to the top side of the printed finger, the magnetic field generator (#1) was placed in front of the finger. The DC motor was controlled by a motor controller (#5, EPOS2 24/2, Maxon Motor). The recorded patient motion data were sent directly to the motor controller using Labview Software (Version 2014, NI®). A conventional PID controller was used to control the system.

To evaluate the ability of the simulator to reproduce tremor torque characteristics, the experimental setup shown on the right side of Fig. 4.1 was developed. The finger model (#7) was disconnected from the motor in order to directly connect the motor shaft to a torque sensor (#4, TRS605-FSH02056, FUTEK). The torque signal was recorded directly onto a PC through a data acquisition card (#3, USB-6002, NI®). The sampling frequency was set to 1 kHz. The nominal torque of the motor with gearhead is 0.5 Nm.
4.1 Development and Validation of a Finger Tremor Simulator

4.1.3.3 Data Processing and Analysis

All data were collected and saved directly onto a personal computer (ASUS® desktop PC M51AC Series, Windows 8). Data processing and analysis were conducted offline using MATLAB (Version R2013a, The Mathworks, Inc.). Prior to data analysis, each signal was digitally low-pass filtered with a 2\textsuperscript{nd}-order Butterworth digital filter with a cutoff frequency of 20 Hz to eliminate high-frequency noise. The quantification of tremor starts with the identification of the onset of the tremor signals. A 1-second window was swept through the entire signal to determine the start and the end of the tremor signal. The data in between these endpoints were used as the tremor signal for the purpose of data analysis and comparison.

4.1.4 Results and Discussion

The assessment of the tremor simulator was divided into two sections: 1) assessment of the reproduction of tremor motion; and 2) assessment of the output torque of the tremor simulator. Sixteen patient data sets were used to complete the assessments.

4.1.4.1 Reproduction of Tremor Motion: Preliminary Testing

To evaluate tremor motion, both tremor and combined motion (tremor and voluntary motion) were reproduced and assessed. Fig. 4.2 shows the real-time comparison of one recorded patient data set (blue curves) and the output from the simulator (red curves). The top figure shows the combined motion and the bottom figure shows the tremor motion. The visual comparison indicates a high correlation between the recorded patient data and the output of the tremor simulator.

Our previous study has shown that parkinsonian finger tremor consists of multiple harmonics, and the second and third harmonics cannot be ignored [3]. In order to assess the reproduction of each harmonic, a Fast Fourier Transform (FFT) was used. The power of the 1\textsuperscript{st}, 2\textsuperscript{nd} and 3\textsuperscript{rd} harmonics of the recorded patient data and the output of tremor simulator is shown in Fig. 4.3. The “TrexH” and “RepxH” represent the x-th harmonic of the recorded patient tremor motion and the simulated tremor motion. The mean power value and standard deviation of each harmonic were calculated from the patient data. The correlation coefficient between the two signals is shown
4.1 Development and Validation of a Finger Tremor Simulator

Figure 4.2: Time series of the recorded patient data (blue) and the output of the simulator (red). The figure at the top shows the combined motion and the figure at the bottom shows the tremor motion.

Figure 4.3: The comparison of the power distribution of the 1st, 2nd and 3rd harmonics. The “TrexH” represents the x-th harmonic of the recorded patient tremor motion and the “RepxH” represents the simulated tremor motion. The correlation coefficient is shown in red. The y axis of the correlation coefficient is labeled at the right side of the figure. In Fig. 4.3 as well, with the y axis shown on the right side of the figure.

The comparison of each harmonic in Fig. 4.3 shows a similar power range between the two signals; however, as the order of the harmonic increases, the power of RepxH deviates more from
4.1 Development and Validation of a Finger Tremor Simulator

Figure 4.4: Distribution of the percentage RMSE and the correlation coefficient for the combined motion. The boxplot on the left shows the percentage RMSE, and the boxplot on the right shows the correlation coefficient.

TrexH. The mean value of the ratios between TrexH and RepxH are 1.09, 1.82 and 2.41 for $x = 1, 2, 3$. This indicates that the tremor simulator has a lower ability to reproduce the higher tremor harmonics. As a result, the correlation coefficient is compromised. To improve these results, further fine tuning of the control system parameters should be performed.

Fig. 4.4 shows the performance of the simulator when reproducing the combined motion. The percentage root mean square error (RMSE) is calculated as the percentage of the error between two signals with respect to the recorded patient data. The mean value ± standard deviation is 11.89% ± 4.01%. The correlation coefficient is 0.99 ± 0.01. Although the acquired percentage RMSE is high, the correlation coefficient shows a good match between the recorded patient data and the output of the tremor simulator. This result is likely caused by under-tuned gains in the control system. In addition, the presence of tremor diminishes as the patient starts to move; thus, while tremor does not completely disappear, the ratio of tremor to the patient’s voluntary motion is reduced. Therefore, the impact of reproducing higher harmonics on the current system becomes insignificant.

4.1.4.2 Output Torque of the Tremor Simulator: Preliminary Testing

Using the data from all 16 patients, the RMS value and the maximum amplitude of the torque of the index finger MCP joint tremor were calculated (Table 4.1). These two indices were used to
4.1 Development and Validation of a Finger Tremor Simulator

Table 4.1: Tremor torque at the index finger MCP joint.

<table>
<thead>
<tr>
<th></th>
<th>RMS ($p &gt; 0.05$)</th>
<th>Maximum Amplitude ($p &gt; 0.05$)</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Patient Tremor Torque</strong></td>
<td>0.02 ± 0.02 Nm</td>
<td>0.06 ± 0.05 Nm</td>
</tr>
<tr>
<td><strong>Simulator Output Torque</strong></td>
<td>0.03 ± 0.01 Nm</td>
<td>0.06 ± 0.03 Nm</td>
</tr>
</tbody>
</table>

Figure 4.5: Time series of the recorded tremor torque from a patient (*blue*) and the output torque of the tremor simulator (*red*).

compare with the simulated torque.

Fig. 4.5 shows the recorded tremor torque from one patient and the output torque of the proposed system. The torque from the simulator was acquired when the patient’s tremor motion data were sent to the controller. An upper limit was set for the current of the motor in the motor controller so that the output torque is limited to 0.2 Nm; therefore, the tremor simulator will not produce torque higher than the real tremor torque. This method allows the simulator to reproduce the tremor motion as close to the real tremor as possible while having the output torque stay within the amplitude range of the real tremor torque. The RMS value of the tremor simulator’s output torque reproducing all patient data is 0.03 ± 0.01 Nm and the maximum amplitude is 0.06 ± 0.03 Nm (Table 4.1). In comparison with the tremor torque from patient data analysis, a $t$ test showed no significant difference. This indicates that the simulated torque has a similar range to the real tremor torque.
4.1.5 Conclusion

The design and validation of a finger tremor simulator was presented in this paper. In addition, the torque of the tremor in the index finger MCP joint of 16 parkinsonian patients was presented. Two preliminary experiments were conducted to validate the performance of the proposed system for reproducing tremor motion and tremor torque. The first experiment on tremor motion reproduction was conducted using all 16 patient data sets, and the second experiment on tremor torque reproduction was conducted using all patient data sets with controlled torque output.

The motion assessment showed that the proposed system is able to reproduce parkinsonian tremor motion, however, the frequency analysis showed that its performance decreases from the lower harmonic to the higher harmonic. This indicates that the control system of the proposed simulator was not optimally tuned. Lastly, the assessment of the proposed system to reproduce combined motion showed a high correlation between the patient tremor motion and the output of the simulator—the percentage RMS error is 11.89% ± 4.01%. The analysis of the patient’s tremor torque at the index finger MCP joint showed that the RMS value is 0.02 ± 0.02 Nm, and the maximum amplitude is 0.06 ± 0.05 Nm. The torque assessment showed that the system is able to produce sufficient torque (RMS value of 0.03 ± 0.01 Nm and maximum amplitude of 0.06 ± 0.03 Nm) and the output torque can be adjusted by changing the limit on the current to the DC motor.

As part of the future work, more patient data will be recorded and used to validate the proposed tremor simulator. In addition, the control system of the proposed simulator will be fine-tuned to improve the performance for reproducing higher harmonics. Lastly, this system will be used to validate a wearable tremor suppression device.

4.2 Development of a Wearable Tremor Suppression Glove

Following the validation of a physical tremor simulator, a wearable tremor suppression glove prototype was developed. In this section, the design of the wtsg prototype and its validation on the physical tremor simulation were introduced.
4.2 Development of a Wearable Tremor Suppression Glove

4.2.1 Abstract

Current treatments for parkinsonian tremor, such as medication and brain surgery, have shown varying levels of effectiveness and carry the risk of significant side effects and complications. Studies on wearable tremor suppression devices have shown positive results in the use of mechanical and electrical suppression on tremor management of the upper limbs. Wearable technology for tremor suppression is a promising solution for patients who do not respond to medication and do not present severe enough symptoms to undergo surgery. Available tremor suppression devices are mainly for elbow and wrist tremor. Devices for finger tremor suppression have not been developed despite the fact that finger tremor is also present. In this study, a wearable tremor suppression glove prototype was designed and validated with recorded tremor data from patients with Parkinson’s disease. Two validation experiments were conducted to assess the performance of the proposed device when suppressing tremor motion and following voluntary motion. The tremor suppression assessment showed an overall tremor amplitude reduction of 85.0% ± 8.1%, and the power reductions for the 1st, 2nd, and 3rd harmonics are 87.9% ± 13.6%, 92.0% ± 7.4%, and 81.7% ± 13.0%, respectively. Following voluntary motion was possible with a RMSE of 14.2% ± 2.5% and a correlation coefficient of 0.97 ± 0.01. Both assessments have shown positive results for the validation of the proposed device; however, further work is needed to improve the performance of the proposed device prior to human trials.

4.2.2 Introduction

Parkinson’s disease is considered to be one of the most prevalent chronic neurological diseases in the elderly [19]. The symptoms of PD are motor-related, including tremor, slowness of movement, rigidity, difficulty with walking, etc. [20]. Parkinsonian tremor is presented as an unintentional, rhythmic, oscillatory movement with a relatively stable frequency and time varying amplitude [1–3]. Clinically, parkinsonian tremor is divided into resting tremor and postural tremor [4]. The reported frequencies of parkinsonian tremor range from 3.5 Hz to 11.5 Hz [3, 5]. Our recent work showed that parkinsonian finger tremor is composed of multiple harmonics rather than a mono-frequency oscillation [3]. Current treatments for PD are medication and brain surgery [6, 7].
4.2 Development of a Wearable Tremor Suppression Glove

Medication is usually effective for slowness and rigidity, but much less effective for treating tremor and often carries significant side effects [21]. Brain surgery is often effective in managing tremor; however, it carries a risk of significant complications [9]. Considering the drawbacks of the traditional treatments, an alternative approach for tremor management has been proposed and studied since Rosen et al. found that mechanical loading can be used to effectively suppress tremor [10]. Externally worn exoskeleton-type therapy devices offer an alternative approach for tremor management, and have demonstrated successful tremor suppression [11, 12, 16, 17]. These devices are a noninvasive option that suppress tremor using mechanical loads generated by actuator(s). Such devices often adopt a tremor estimator in order to distinguish the tremor motion from the voluntary motion. Force, position, and electromyography (EMG) signals are the most commonly used inputs for controlling device movement. Electrical suppression of tremor motion has also been proposed [8, 13, 22, 23]. This approach counterbalances the contraction of the tremorous muscles by sending electrical current to the antagonistic muscles. The developed devices are lighter and more compact than mechanical suppression devices, but they can cause discomfort, pain, and muscle fatigue, while resulting in lower suppression performance. Tremor in the elbow and the wrist have been studied extensively [8, 11–13, 16, 17, 22, 23], but finger tremor has not yet been well studied, despite the fact that finger tremor is also present. Moreover, our recent study showed that it is not sufficient to suppress only the tremor in the proximal joints, as the tremor magnitude in the distal joints increases when the tremor in the proximal joints is suppressed [24]. Among most of the aforementioned devices, mechanical suppression devices achieved higher suppression than the electrical suppression devices. In particular, the Wearable Orthosis for Tremor Assessment and Suppression (WOTAS) showed superior performance, with a 90% suppression rate for tremor amplitude [11]. Although mechanical suppression has superior suppression performance, the size, weight, and power consumption of such devices remain as significant challenges to be addressed. Furthermore, no finger tremor suppression device has been developed. Therefore, the goal of this work was to develop and validate a compact wearable tremor suppression glove (WTSG) that can be used to mechanically suppress parkinsonian tremor of the hand without affecting the voluntary motion. The WTSG was designed to suppress tremor in the index finger metacarpophalangeal (MCP) joint, thumb MCP joint, and wrist in the flexion-extension direction. Presumably, reduced
tremor in these joints will lead to some practical benefit through the provision of stable lateral or precision pinching, for instance; however, this is not explored in this study. The WTSG was validated on a tremor simulator with recorded tremor data from patients with PD.

4.2.3 Device Design and Construction

In this section, the mechanical design, sensing, actuation, and control system of the WTSG are presented.

4.2.3.1 Prototyping of the WTSG

The WTSG consists of a sensing glove, an actuation block, and a cable-enabled power transmission system. The cable-enabled transmission system forms the bridge between the other two sections of the WTSG. It uses non-stretchable flexible cables to transmit the pulling force to the target joint; the flexion and extension motions of each joint are controlled by the actuation block through two cables. Based on the conceptual WTSG model introduced above, a CAD model was created (Fig. 4.6). The frame of the WTSG includes two finger rings (1 and 2), two hand supports (palm side (3) and dorsal (4) side), one thumb support (5), one wrist support (6), and one rotary-to-linear converter (7). The cables $e_t$ and $e_t$ for index finger and thumb extension, respectively, are fixed at the top edge of the finger and thumb rings through the guide tubes ($t_1$ and $t_3$) on the hand support (4) and thumb support (5), and the cable $e_w$ for wrist extension is fixed at the edge of the hand support (4). The cables $f_f$ and $f_w$ for index finger and wrist flexion are fixed at the bottom edge of the finger ring and the edge of the hand support (3). The guide tubes ($t_4$) on the hand support (3) facilitate the movement of the cable $f_t$. The wrist support (6) changes the movement direction of cables $f_f$ and $f_w$. Lastly, the cable $f_t$ for thumb flexion is fixed at the bottom edge of the thumb ring through the guide tubes ($t_2$) on the thumb support (5).

The details of the wrist support are shown in Fig. 4.7. The main function of the wrist support is to align the cables $f_f$ and $f_w$ to the center axes of the hand and index finger for proper power transmission. The placement of cables $f_f$ and $f_w$ are supported with three guides ($#13$, $#14$, and $#15$) fixed in the wrist support. These guides utilize bearings with an extended inner ring to create guide slots for the cables. Since the outer rings are fixed to the structure, the inner rings can move
4.2 Development of a Wearable Tremor Suppression Glove

Figure 4.6: CAD model of the WTSG. The red lines and green lines represent the non-stretchable cables on the dorsal and palm side of the hand. 1 and 2. finger ring, 3 and 4. hand support, 5. thumb support, 6. wrist support, 7. rotary-to-linear converter

Figure 4.7: Configuration of the wrist support and cable wiring method. a) top view. b) bottom view. c) isometric view. d) side view. e) front view. 1–3 are DC motors, 4, 10, and 19 are double groove pulleys, 5, 6, 8, 9, 11, 16, 18, and 20 are ball bearings, 7, 12, and 17 are stainless steel pins, 13, 14, and 15 are bearing guides. Cables in green control the flexion of the joints; cables in red control the extension of the joints. Cables in green control the flexion of the joints; cables in red control the extension of the joints.

freely with the cables.

The details of the rotary-to-linear converter are also shown in Fig. 4.7. Its main function is to transfer the rotary motion of the DC motors to the linear motion of the non-stretchable cables.
4.2 Development of a Wearable Tremor Suppression Glove

Figure 4.8: (a) Configuration of the double groove pulley. (b–d) Prototype of the wrist support. The dimensions of the wrist support are indicated in the boxes. Units: mm

The front view of the converter is shown in Fig. 4.7 (e). The DC motors (1), (2), and (3) control the motion of thumb, index finger, and wrist through cables \( f_t \), \( f_i \), \( f_w \), \( e_t \), \( e_i \), and \( e_w \), Fig. 4.7 (a). These cables are fixed to two double groove pulleys (4, 10, and 19) that are press fitted onto the shafts of the motors. The cable winding method for the pulley is shown in Fig. 4.8 (a). Two through holes (black region in Fig. 4.8 (a), \( left \)) are located at the bottom of each groove, their longitudinal axes are perpendicular to the flat surfaces of the motor shafts. The non-stretchable cable insertion direction is depicted at the right side of Fig. 4.8 (a). After the insertion, the sections of the cable in the through holes were affixed with glue. The loose ends of the cable were wound in opposite directions in each groove (Fig. 4.8 (a), \( left \)). To guide and keep the cables in position, eight ball bearings (5, 6, 8, 9, 11, 16, 18, 20) were used. Stainless steel pins (7, 12, and 17) function as simple direction changing mechanisms, with two ball bearings installed on each end to reduce the friction between the cables and the pins.

Based on the CAD model, a prototype was created as shown in Fig. 4.8 (b–d). The height,
width, and length of the prototype are 66.3 mm, 86.8 mm, and 29 mm, respectively. The total weight of the current prototype is 300 g. This weight includes three DC motors, but it does not include the finger rings and hand supports, as they are custom made for each person. An approximated range of the weight of these attachments is 20 to 40 g. Since mechanical loading has been shown to suppress tremor effectively [10], the weight of the proposed device is likely to affect the tremor motion; however, the quantification of such impact is not investigated in this study due to the nature of the current benchtop experimental setup.

The WTSG prototype is shown in Fig. 4.9. The finger rings and hand supports were manufactured in blue plastic, the thumb support in white, and the wrist support in pink.

4.2.3.2 Sensing, Actuation and Control System

The sensing system of the WTSG consists of five inertial measurement units (LSM9DS1, STMicroelectronics®). Each IMU consists of one 3 degree-of-freedom (DOF) accelerometer, one 3 DOF gyroscope, and one 3 DOF magnetometer. The hardware of the sensing system and the location of each IMU are shown in Fig. 4.10. IMU 1 and the microcontroller (MCU) are attached to the wrist support, IMUs 2, 3, 4, and 5 are attached to the thumb support, hand support (dorsal
4.2 Development of a Wearable Tremor Suppression Glove

Figure 4.10: Configuration and location of the IMUs and the microcontroller.

side), thumb ring, and index finger ring, respectively. The motion of each target joint is measured differentially with two nearby IMUs. All sensors communicate with one microcontroller (NXP® LPC1768) through a serial peripheral interface (SPI) at a sampling rate of 100 Hz.

The actuation system consists of three brushless DC (BLDC) motors and one cable-actuated power transmission system. Each motor is controlled directly by a motor controller (Maxon Motor® EPOS2 24/2). All three motor controllers are controlled by the microcontroller (MCU) at the same time. Two Maxon Motor® EC-max 16 BLDC motors with planetary gearheads (29:1) were used for the finger and thumb, and one Maxon Motor® EC 16 BLDC motor with planetary gearhead (29:1) was used for the wrist.

In order to control the motion of the patients’ fingers and wrist, the voluntary signal needs to be extracted from the measured signal. The performance of the control system is highly dependent on the performance of the tremor estimator. Therefore, it is necessary to estimate tremor with minimal error.

The flow diagram of the control system is shown in Fig. 4.11 The measured signal is sent to the control system and is processed by the tremor estimator. The voluntary motion estimator subtracts the estimated tremor signal from the measured signal, and then sends the estimated voluntary motion to the DC motor. The motor controls the cables to move according to the
estimated voluntary motion, while applying enough force to suppress the tremor. This method of control reduces the requirements on the actuators since the movement of the voluntary motion is slower than the tremor.

In practice, joint position data are calculated from the angular velocity measured through IMUs. Since the angular velocity does not directly reflect the joint angle, an integration algorithm is applied at the beginning of each control cycle to calculate the joint angle.

As mentioned above, the proposed control strategy is to control the devices following the voluntary movements of the user. Therefore, the tremor component needs to be eliminated from the measured signal before it is sent to the DC motors. A number of algorithms have been proposed to estimate tremor, such as Fourier Linear Combiner (FLC) [25], Weighted-frequency Fourier Linear Combiner (WFLC) and its derivatives [26, 27], and Kalman filter based estimators [28]. Although more complicated algorithms have shown better estimation accuracy, in this study, a WFLC was adopted as the tremor estimator due to its lower computational cost.

4.2.3.3 Tremor Characteristics and Tremor Simulator

The tremor torque in the index finger MCP joint in the flexion-extension direction was assessed in our previous study [29]. The root mean square (RMS) value of the tremor torque at the MCP joint is $0.02 \pm 0.02$ Nm, and the maximum amplitude is $0.06 \pm 0.05$ Nm.

To reproduce the real tremor, the maximum tremor torque was set as the output limit of
the BLDC motor of the tremor simulator. The recorded patient data were sent directly to the
tremor simulator for validating the WTSG. For each patient data set, two types of motion were
reproduced. The first one is tremor motion, which was recorded when the patient’s hand was
at rest; the other one is tremorous voluntary motion, which was recorded when the patient was
performing an oscillatory motion with their hand.

4.2.4 Experimental Evaluation

4.2.4.1 Experimental Setup for WTSG Validation

The experimental setup for the validation of the WTSG is shown in Fig. 4.12 [29]. The setup
consists of a tremor simulator, the WTSG, and an electromagnetic (EM) motion tracking system
(Aurora, NDI®). In this preliminary evaluation, only the index finger module of the WTSG was
validated on the tremor simulator. The simulator is controlled directly by a PC using a customized
user interface (Labview Software Version 2014, NI®). The WTSG was securely fixed onto a table.
The index finger ring was placed onto a 3D printed finger, and two dowel pins were placed in
between the 3D printed finger and the WTSG to guide the non-stretchable lines for index finger
flexion and extension. IMU 5 was fixed on the top of the index finger ring.

In addition to the IMU, a 5 DOF EM motion sensor was also attached onto the index finger ring.
The magnetic field generator was placed in front of the printed finger. The angular displacement
of the printed finger was recorded directly in the PC through the Aurora’s customized software
interface. The sampling frequency was configured to the maximum value of 40 Hz.

4.2.4.2 Data Processing and Analysis

All data from the validation experiment were recorded and stored directly onto a PC (ASUS®
desktop PC M51AC Series, Windows 8). Data processing and analysis were performed offline using
MATLAB software (Version R2013a, The MathWorks, Inc.). The percentage power reduction and
the RMS error were used to quantify the performance of the WTSG suppressing the tremor and
following the voluntary motion. A one second time window was swept through the data set to
determine the beginning and the end of the tremor. The data in between were used for the
4.2 Development of a Wearable Tremor Suppression Glove

Figure 4.12: Experimental setup for assessing tremor suppression in a single finger. The tremor simulator is shown in the black box, and the WTSG is shown in the red box. The Aurora magnetic field generator was placed in front of the tremor simulator.

calculation.

In the following sections, two performance assessments of the WTSG are presented: 1) assessment of tremor suppression; and 2) assessment of performance when following voluntary motion. Seven patient data sets were used to complete the assessments.

4.2.4.3 Performance of the Tremor Suppression

Fig. 4.13 shows the comparison between the patient tremor signal (blue curves) and the suppressed motion of the printed finger (red curves) from one trial. The figure at the top shows the angular displacement of the two signals as a function of time. It is clear from this comparison that the majority of the tremor motion was reduced by the WTSG. A similar result was obtained from the frequency domain analysis. The figure at the bottom presents the power distribution of the two signals. The patient tremor data present multiple harmonics with the fundamental frequency located at about 5 Hz. In comparison, the motion of the printed finger shows large power reductions at the 1st, 2nd, and 3rd tremor harmonics.

Our previous study showed that parkinsonian finger tremor presents multiple harmonics, and
4.2 Development of a Wearable Tremor Suppression Glove

Figure 4.13: One patient tremor data (blue) and the suppressed motion of the simulator (red) in both time (top) and frequency domain (bottom).

that the 2nd and 3rd harmonics cannot be neglected [3]. To evaluate the tremor reduction of each harmonic, the percentage power reduction over all patient data sets is shown in Fig. 4.14. The “Hx” represents the x-th harmonic. Both the mean value and standard deviation are given. The percentage reduction of the tremor displacement amplitude (“Overall”) is also shown in Fig. 4.14.

The overall tremor amplitude reduction (mean ± standard deviation) of the WTSG is 85.0% ± 8.1%, and the power reduction for the 1st, 2nd, and 3rd harmonics are 87.9% ± 13.6%, 92.0% ± 7.4%, and 81.7% ± 13.0%, respectively. The mean value of the amplitude reduction is 5% lower than the result of a five-year European project for elbow tremor suppression “Dynamically Responsive Intervention for Tremor Suppression (DRIFT)” [11, 30]. The difference in tremor reduction with the DRIFT study is likely caused by the different validation approaches. Since the tremor simulator used in the current study is a rigid body, it does not present the level of compliance of a human finger. In addition, other possible reasons for the lower performance include insufficient tension in the non-stretchable cable, which results in relative movement between the index finger ring and the printed finger; the fact that the WTSG transmission system did not include an idler to compensate for the varying travel distance between the flexion and extension cables caused by
the irregular shape of the MCP joint; and the tremor estimator of the WTSG.

4.2.4.4 Performance when Following Voluntary Motion

Although tremor suppression is the main function of the WTSG, the ability to follow voluntary motion should also be evaluated because the WTSG will be worn when completing everyday tasks. Fig. 4.15 shows the performance of the WTSG (red curves) when following the voluntary motion from one patient data set (blue curves). The visual comparison of the signals in the time domain indicates a high correlation (Fig. 4.15, top). The comparison in the frequency domain is shown in Fig. 4.15, bottom. Similar values were obtained from both signals at a frequency lower than the tremor bandwidth. The inset in Fig. 4.15 shows the power of both signals at the tremor frequency range. The data from the printed finger have lower power than the patient data. This result clearly shows that the tremor component was suppressed by the WTSG with only a small impact on the voluntary motion.

The percentage RMSE (14.24% ± 2.52%) and the correlation (0.97 ± 0.01) from all seven trials are shown in Fig. 4.16. Since it is impossible to compare these results with other finger tremor suppression devices, as they do not exist, the result of a recent study validating a finger exoskeleton device on an able-bodied subject was used for comparison [31]. This study reported a
4.2 Development of a Wearable Tremor Suppression Glove

4.2.5 Conclusions and Future Work

In this paper, the design process and experimental validation of a WTSG was presented. The experimental validation was conducted on a tremor simulator with seven recorded patient data sets. Two experiments were conducted for each trial, including the assessment of tremor suppression and the assessment of the ability to follow voluntary motion. The tremor suppression assessment showed an overall tremor amplitude reduction of $85.0\% \pm 8.1\%$, and a power reduction for the 1st, 2nd, and 3rd harmonics of $87.9\% \pm 13.6\%$, $92.0\% \pm 7.4\%$, and $81.7\% \pm 13.0\%$, respectively. The performance assessment when following the voluntary motion showed that the percentage RMSE is $14.2\% \pm 2.5\%$ and the correlation coefficient is $0.97 \pm 0.01$. 

Figure 4.15: Voluntary motion from one patient (blue) and the output of the WTSG following the voluntary motion (red) in both time domain (*top*) and frequency domain (*bottom*). The inset in the bottom figure shows a lower tremor power in the output of the WTSG (red) than the patient data (blue).

12\% error in the voluntary motion prediction at the MCP joint. Although the percentage RMSE of the proposed WTSG is not ideal, the correlation coefficient shows a good match between the two signals. Similarly, the reasons for a relatively high RMSE may be same as those discussed for the suppression of tremor being lower than desired.
4.2 Development of a Wearable Tremor Suppression Glove

Figure 4.16: Distribution of the percentage RMSE and the correlation coefficient from seven trials. The boxplot on the left shows the percentage RMSE, and the boxplot on the right shows the correlation coefficient.

Both assessments have shown positive results for the validation of the functions of the WTSG. However, possible issues that may affect the performance of the WTSG were also identified. Future work will use more patient data to assess the WTSG after the identified problems have been addressed to improve the performance of the WTSG. In addition, a whole upper limb tremor suppression device will be developed and validated. Lastly, an evaluation of the WTSG on patients will be conducted.
Bibliography


Chapter 5

Second Generation WTSG and Pre-clinical Validation

This chapter is adapted from ‘An Alternative Approach for Pathological Tremor Management – A Wearable Tremor Suppression Glove: Design and Validation’, submitted to IEEE Transactions on Mechatronics, 2019.

5.1 Abstract

Pathological tremor is characterized by the unintentionally reciprocating motion of a part of the human body. The presence of tremor significantly reduces the quality of life of the individuals who are living with tremor, creating difficulties for fine motor tasks, such as eating and writing, and causing social embarrassment. Current treatments include medication and brain surgery, however, they both are associated with a number of side effects and complications. The emergence of wearable technology facilitates an externally worn exoskeleton-type tremor suppression device as a potential alternative approach for tremor management. This paper details the development of a wearable tremor suppression glove for suppressing tremor in the index finger metacarpophalangeal (MCP) joint, thumb MCP joint and the wrist, as well as the validation of the proposed device with one subject living with parkinsonian tremor. Results from the experiment showed an overall suppression of 73.1%, 80.7% and 85.5% in resting tremor, 70.2%, 79.5% and 81% in postural
5.2 Introduction

Parkinson’s disease (PD) is considered to be the second most common neurological disease after Alzheimer’s [1]. The overall numbers are growing each year and are estimated to double in the next 15 years in both developing and developed countries [2–5]. As worldwide life span has increased, the progressive and chronic nature of PD has led to a substantial economic burden on health care systems and individual patients [4, 6, 7]. Tremor is one of the most severe motor-related symptoms of PD. It often appears as an involuntary, rhythmic, oscillatory movement with a time-varying magnitude and relatively stable frequency [8]. A recent study has shown that parkinsonian tremor consists of multiple harmonics, and that the second and third harmonics cannot be neglected [9]. Clinically, two types of tremor can be found in people with PD, one is the resting tremor and the other is the postural tremor [10]. The reported fundamental frequency of parkinsonian tremor ranges from 3.5 Hz to 7.7 Hz, and the magnitude of tremor increases from the proximal joints to the distal joints [11].

Current treatments for PD include both medication and brain surgery [12, 13]. Medication, being the mainstay of treatment, is usually less effective for treating tremor than rigidity and slowness, and often presents side effects and issues with drug tolerance [14, 15]. In case of severe tremor, brain surgery is considered; however, this treatment is invasive in nature and carries a significant risk of complications for the individual [16]. In addition to the traditional treatments, the emergence of wearable technology, for example [17–19], facilitates an externally worn exoskeleton-type device as a potential alternative approach for managing tremor.

Adelstein et al. discovered that tremor can be effectively suppressed by applying mechanical loading [20], which has encouraged researchers to develop externally worn exoskeleton-type mechatronic devices for tremor management. Since then, several research groups have developed tremor suppression devices for both wrist and elbow tremor [21–37]. These devices were designed to use...
electromechanical actuators to generate controllable force to selectively suppress the tremor in the wrist and elbow. Their control system often includes a tremor estimator that extracts the tremor signal from the measured signal, which can be force, position, inertia, and/or electromyography (EMG) signals.

In most studies, the validation of these proposed devices was performed with simulation, in non-wearing tests using a benchtop tremor simulator, or with healthy volunteers [22–29, 32, 33, 35–37]. Although these studies have shown a promising ability to suppress tremor, they still need to be validated on human subjects with PD to facilitate acceptance by clinicians and patients.

In the scope of tremor suppression devices, several devices have been designed and evaluated on human subjects with PD [21, 30, 31, 34]. These devices not only showed encouraging performance on tremor suppression, but also helped the transition of externally worn exoskeleton-type tremor suppression devices from lab evaluation to human validation. Lastly, putting aside the differences between these studies, one noticeable common ground is that they all focused on tremor in the wrist and in the elbow.

A recent study has raised an interesting question regarding the tremor management approach [38]. Does the suppression of tremor in only the proximal joints meet the requirements of daily use for the end user? This study found that applying mechanical suppression to the proximal joints, such as the elbow and the wrist, results in an increase in the tremor magnitude in the distal joints, such as the fingers. As most of the activities of daily living involve the use of fingers, for the user to benefit from externally worn exoskeleton-type mechatronic devices, the suppression of finger tremor must be considered.

To address this issue, several studies have focused on the development and validation of externally worn exoskeleton-type tremor suppression gloves [39, 40] for independent finger tremor suppression. Similar to most of the aforementioned studies, these studies achieved good tremor suppression performance on lab simulation setups. However, none of them has been validated on human subjects. Therefore, the goal of the current study is to develop a wearable tremor suppression glove (WTSG) and to validate it on a human subject with parkinsonian tremor.

In the framework of the alternative approach for pathological tremor management project, the WTSG was designed to measure, extract, and suppress tremor in the finger joints and wrist
while allowing the user to perform voluntary motion with minimum impact from the suppression. This paper presents the development of the WTSG and its validation on one human subject with parkinsonian tremor.

5.3 Mechatronic Design

5.3.1 Conceptual Model and Prototyping

The conceptual 3D computer-aided design (CAD) model was developed using SOLIDWORKS, as shown in Fig. 5.1 and Fig. 5.2. The exoskeleton frame of the WTSG consists of eight pieces, including two finger rings (Fig. 5.2, F and G) located at the index finger proximal phalange and thumb proximal phalange, one dorsal plate (Fig. 5.2, E) located at the back of the hand, one palm plate (Fig. 5.1, D) located at the palm, one thumb plate (Fig. 5.2, H) and one sheath holder (Fig. 5.2, I) located at the thumb metacarpal, and two actuation blocks (Fig. 5.1, A and B) located on the forearm. The electronic and actuation systems of the WTSG are built on top of the exoskeleton frame, they consist of five inertial measurement units (Fig. 5.1, #2 and #7), one microcontroller (Fig. 5.1, #7), three brushless DC motors (Fig. 5.1, #1), three motor controllers (Fig. 5.1, #6), and three double grooved pulley-cable units (Fig. 5.1, #4 and #5). The goal of the current design is to suppress tremors in the index finger MCP joint, thumb MCP joint, and the wrist in the direction of flexion and extension.

The existence of the exoskeleton frame not only provides a physical support to the electronic and actuation systems, but also plays an important role in motion and force transmission from the actuation system to the target joints. To minimize the size, and to adapt the WTSG to the user’s hand, a pulley–cable transmission system was adopted (Fig. 5.3). This transmission system is superior to other types of transmissions for remote actuation, as required by wearable devices. In the current design, the pulley–cable transmission system consists of three components—the double grooved pulley (Fig. 5.3, #1), the flexible steel cable (Fig. 5.3, #3), and the spiral cable slewing (Fig. 5.3, #2). The pulley was designed to have two grooves, indicated as “a” and “b” in Fig. 5.3, Top View, for the cable to be wound such that the two ends of a cable retract and release at the same time as the motor rotates the pulley. This arrangement allows the WTSG to use one
Figure 5.1: CAD model of the WearME Glove. The red lines and green lines represent the cable sleeving used to guide the soft steel cables (blue lines) for joint flexion and extension. 1. Brushless motors, 2. Inertial measurement units (IMUs), 3. Setscrews, 4. Double grooved pulley, 5. Soft steel cable, 6. Motor controllers, 7. Main controller. *. Structural support for cable sleeving, +. Fixture structure for cable termination.

motor to remotely control both flexion and extension of a target joint. The cable winding method is shown in Fig. 5.3, Front View. There is one through hole at the bottom of each groove to form a path for the cable to go through. The insertion direction of the cable is depicted with two arrows in the figure. After the insertion, both loose ends of the cable were wound in opposite directions (Dir. 1 and Dir. 2 in Fig. 5.3) in each groove. To successfully transfer the motion and force to the target joint, spiral cable sleevings were placed on the exoskeleton frame to provide soft guidance to the cable. The cable sleevings for joint extension and flexion motions are shown as the green
5.3 Mechatronic Design

Figure 5.2: CAD model of the WearME Glove. 3. Setscrew. * Structural support for cable sleeving, +. Fixture structure for cable termination.

Figure 5.3: Pulley–cable transmission system. 1. Double grooved pulley, 2. Cable sleeving, 3. Soft steel cable. Top view of the pulley shows both grooves of the pulley, indicated by "a" and "b". Front view of the pulley shows the winding direction of a cable onto the pulley. The diameter of the pulley is 8 mm.
and red curves in Fig. 5.1 and Fig. 5.2. The steel cables are shown as blue curves. To secure the cable sleeving, several structural supports, which are labeled as “*” in Fig. 5.1 and Fig. 5.2, were used in the current design.

To effectively transfer motion and force to the target joint, the terminal ends of each cable must be securely fixed to the exoskeleton frames, which are Piece D and Piece E for the flexion and extension of the wrist, Piece F for the flexion and extension of the index finger MCP joint, and Piece G for the flexion and extension of the thumb MCP joint. Each terminal end passes through the channel of a fixture structure, which is labeled as “+” in Fig. 5.1 and Fig. 5.2, on each exoskeleton frame. Set screws (Fig. 5.1 and Fig. 5.2, #3) were used to secure the cable to the fixture structure by applying direct pressure onto the cable.

To determine the minimum number of set screws to use, a benchtop loading test was performed (Fig. 5.4). The required stall force ($F_s$) was determined by the tremor torque ($\tau_t$), the distance between the fixture structure and the joint ($d$), and the maximum joint angle ($\theta$) using the following
5.3 Mechatronic Design

Figure 5.5: The WTSG prototype. 1, 2 and 3. Velcro straps, a, b, c, d. Location of the IMUs, e. Location of the main controller. Total weight of the WTSG prototype is 450 grams.

equation:

\[ F_s = \frac{\tau_t}{d \cdot \tan(90^\circ - \theta)} \]  \hspace{1cm} (5.1)

The tremor torque was 0.2 Nm [41], the average distance between the fixture structure and the joint was 15 mm, and the maximum joint angle was 70°. Therefore, the required stall force should be at least 36.6 N. The test was conducted on a steel cable with a rated capacity of 10 lbs. The loading test showed that two set screws could hold up to 6 kg (13.2 lbs) of weight before the cable snapped. This capacity is sufficient for effective motion and force transmission.

Based on the CAD model, a prototype was created as shown in Fig. 5.5. The 3D printed pieces were sewn onto a compression glove. Three elastic Velcro straps (Fig. 5.5, #1, #2 and #3) were sewn onto the glove to secure the glove and exoskeleton frame to the user’s hand. This integration reduces the motion disturbance from the cable to the exoskeleton frame when the DC motors are
activated. The total weight of the WTSG is 580 g (1.3 lb). This weight includes three DC motors, three motor controllers, the main controller, and the inertial measurement units (IMUs).

### 5.3.2 Sensing, Actuation and Control System

The measurement system (Fig. 5.6) of the WTSG includes five IMUs (LSM9DS1, STMicroelectronics®). Each IMU has one 3 degree-of-freedom (DOF) gyroscope, one 3 DOF accelerometer, and one 3 DOF magnetometer. Four IMUs were placed in the sensor slots of four exoskeleton frame pieces (Fig. 5.5, a, b, c, and d). The fifth IMU was placed on the main controller board located below the wrist (Fig. 5.5, e). The angular velocity of each target joint was measured differentially with two nearby IMUs. The angular velocity data from all IMUs were acquired by the main controller (NXP® LPC1768) through SPI at a sampling frequency of 70 Hz.

The actuation system consists of one Maxon Motor® EC-max 16 BLDC motor with a planetary gearhead (29:1) for wrist tremor suppression, and two Maxon Motor® EC-max 16 BLDC motors with planetary gearheads (29:1) for index finger and thumb tremor suppression. All three motors are controlled individually by three motor controllers (Maxon Motor® EPOS2 24/2). Each motor controller receives commands directly from the main controller through UART communication at a baud rate of 125000.
5.3 Mechatronic Design

Figure 5.7: Control flow diagram of the WTSG. $u$ represents the measured motion signal, $t$ represents the estimated tremor signal, $y$ represents the estimated voluntary signal.

Lastly, the control system design requires the WTSG to suppress the user's tremor motion while having minimal or no impact on their voluntary motion. Currently, there are two types of tremor suppression strategies: one is to control the WTSG to move antagonistically to the tremor motion, the other is to follow the user’s voluntary motion. The former control strategy relies on estimating the tremor motion component before suppressing it. Such an approach requires the actuation system to move at a high frequency, which may cause physical discomfort to some users. The latter strategy, on the other hand, considers tremor as a disturbance to the system, such that the WTSG traces the user’s voluntary motion while suppressing the tremor component. As mentioned above, the control system of the proposed WTSG adopted the second control strategy.

The flow diagram of the WTSG is shown in Fig. 5.7. The joint’s angular velocity ($u$) measured from the IMUs is first sent to a tremor estimator to determine the estimated tremor motion ($t$). This estimated tremor signal is then subtracted from the original signal ($u$) to generate an estimated voluntary velocity signal ($y$). To estimate the tremor signal, the proposed WTSG adopted the Weighted-frequency Fourier Linear Combiner [42] as the tremor estimator due to its lower computational cost. Lastly, the estimated voluntary velocity was integrated to obtain displacement, and used as the direct input command for the motor controller.
5.4 Experimental Platform and Protocol

In contrast to our initial evaluation of the WTSG on a simulator [40, 43], the evaluation of the current WTSG prototype on a human subject requires a systematic experimental protocol with comprehensive safety measures. In this section, the structure of the safety measures, experimental protocol, and data analysis methods are presented.

5.4.1 Safety Measures and Experimental Setup

The safety measures adopted in the current WTSG are in a three-layer hierarchy, which consists of the software level at the bottom, the hardware level at the middle, and the user level at the top (Fig. 5.8). The software level has the lowest effectiveness, followed by the hardware level, and the user level.

There are three limits configured at the software level: the maximum motor output speed, the maximum motor output torque, and the maximum output position. Since the control strategy requires the motors to follow the voluntary motion of the joints, the voluntary motion of the subject is represented as a sinusoidal signal with a maximum frequency of 2 Hz and a maximum amplitude based on the range of motion (Table 5.1). Therefore, the maximum voluntary motion
Table 5.1: Typical Range of Motion of the Relative Joints and the Motor position Limits

<table>
<thead>
<tr>
<th>Joint</th>
<th>IF Motion</th>
<th>T Motion</th>
<th>W Motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Angle</td>
<td>Ext. 30°</td>
<td>Ext. 15°</td>
<td>Ext. 30°</td>
</tr>
<tr>
<td></td>
<td>Flx. 70°</td>
<td>Flx. 30°</td>
<td>Flx. 50°</td>
</tr>
<tr>
<td>Half P-P Value</td>
<td>50°</td>
<td>22.5°</td>
<td>40°</td>
</tr>
<tr>
<td>Motor Pos. Limit</td>
<td>113°</td>
<td>263°</td>
<td>112°</td>
</tr>
</tbody>
</table>

1. The Half P-P Value is the mid-value between the extension and flexion range of motion.

speeds of the index finger, thumb, and wrist are 105 rpm, 47 rpm, and 83 rpm, respectively. Since the fixture points of the cable terminals are about 15 mm from the joints, and the radius of the double grooved pulleys is 4 mm, the corresponding maximum motor output speeds are 394 rpm, 176 rpm, and 311 rpm.

Our previous study on tremor characteristics [43] has shown that the RMS tremor torque in the fingers is 0.02 ± 0.02 Nm. Based on the configuration of the current WTSG, a 0.02 Nm maximum output torque limit was set to the corresponding motor controllers. In addition to the finger tremor, Belda et al. reported that the RMS tremor torque in wrist is 0.2 Nm [41]. Therefore, a 0.2 Nm maximum torque limit was set for the motor controller that controls the motion of the wrist.

The maximum motor output position was determined by the range of motion of the subject’s target joints (Table 5.1). Based on the configuration of the current WTSG, the maximum output positions of the motors are shown in the last row in Table 5.1.

Above the software level, the hardware level includes two measures: the mini mechanical stopper and the external mechanical stopper. The mini mechanical stopper consists of one 3D printed cube and one setscrew (Fig. 5.9, inset). A steel cable (blue curve in Fig. 5.9) passes through the cube and is fastened by the setscrew. The use of mini mechanical stoppers is mainly for preventing super-extension of the index finger and thumb MCP joints by the WTSG. Their placements are indicated as “m” in Fig. 5.5. To test the resistance capability of these stoppers, a weight test was conducted on a table vice, as shown in Fig. 5.9. An average maximum applied weight of 3.6 kg was obtained over 10 trials. Such capacity is sufficient to stop the corresponding motors with a maximum torque output (i.e., 0.24 Nm).
5.4 Experimental Platform and Protocol

Figure 5.9: Weight test on a mini mechanical stopper. A total weight of 3.6 kg was applied to the stopper in the photo. The CAD model of the stopper is shown in the inset.

In addition to the mini mechanical stopper, an external mechanical frame was developed to limit the subject’s hand motion (Fig. 5.10.b). Two separate sections constitute the mechanical frame for limiting hand extension (Fig. 5.10, ms1) and hand flexion motion (Fig. 5.10, ms2). The allowed range of motion within the mechanical frame is $50^\circ$ for extension and $70^\circ$ for flexion.

The last safety measure with the highest effectiveness is at the user level. In this experiment, two emergency stop buttons were used in serial with the output of the power supply. The subject held one emergency stop, while the other one was held by one of the experimental proctors. Both the subject and the experimental proctor can cut the power of the motors if the WTSG acts improperly.

Lastly, the experimental setup is shown in Fig. 5.10. Fig. 5.10.a presents the subject wearing the WTSG prototype. Three motor controllers (Fig. 5.10, #1) were located on the forearm close to the elbow. The main controller (Fig. 5.10, #2) was placed on the forearm close to the wrist. Each of the IMUs was placed in the corresponding sensor slot with a sensor cover (Fig. 5.10, #3) on top that functions as a strain relief for the wires of the IMU. During the trial, the subject was
Figure 5.10: Experimental Setup. a. The WTSG prototype on the subject’s hand. 1. motor controllers, 2. main controller, 3. IMUs. b. Experimental setup for the validation of the WTSG. ms1 and ms2 are the external mechanical stoppers.

sitting on a chair while the right arm was securely strapped onto a table, allowing only the hand to move freely. This configuration allows the subject’s hand to move only within the scope of the mechanical frame, such that the mechanical frame could provide effective safety to the subject.

5.4.2 Experimental Protocol and User

In order to evaluate the feasibility of the proposed WTSG to suppress tremor, we conducted an experiment on one subject living with parkinsonian tremor. The subject was diagnosed and recruited by a neurologist. The subject’s right hand was selected for the experiment as it presented the greatest amount of tremor. The subject signed a consent form to participate in the study. The experimental protocol was approved by the University of Western Ontario’s Human Research Ethics Board.

The trial was divided into two phases: the tremor monitoring section and tremor suppression phase. During the tremor monitoring phase, only the measurement system (Fig. 5.6) of the WTSG
5.4 Experimental Platform and Protocol

was attached to the subject’s hand. The purpose of this phase is to acquire the unsuppressed tremor from the subject as a baseline for evaluating the performance of the WTSG. During the tremor suppression phase, the subject was wearing the proposed WTSG to perform a series of tasks for evaluating the performance of the WTSG when suppressing different types of tremor. Three different tasks were selected for the subject to perform during each trial:

- **Hand Rest**: maintain the arm in the rest position on a table for 60 seconds.

- **Hand outstretched**: keep the hand outstretched approximately $45^\circ$ above the table level for 60 seconds.

- **Move an object**: pinch a lightweight object with the index finger and thumb, then move it from the hand outstretched position to the rest position for five repetitions.

The aim in selecting these tasks was to acquire different types of tremor, i.e., resting tremor, postural tremor, and kinetic tremor.

While the subject was performing the tasks, the subject was asked to solve simple math questions to distract attention from the desire to suppress the tremor. During the trial, the subject was sitting on a chair while the proctor used customized software to record data from the WTSG and video images of the subject’s hand. No personal identifiers were collected as part of the experiment.

5.4.3 Data Processing and Analysis

The angular velocity data from the IMUs were recorded directly onto a personal computer (ASUS® desktop PC M51AC Series, Windows 8.1) through serial communication using customized software. The software was designed to manage the data transfer and storage from all of the IMUs to the PC with a synchronized timestamp at 70 Hz.

Data analysis was performed offline using MATLAB software (Version R2018a, The Mathworks, Inc.). According to the results from our previous study on hand tremor frequency characteristics [9], a 4th-order Butterworth zero-phase low-pass filter with a cutoff frequency of 30 Hz was first used to eliminate noise while still passing all of the meaningful harmonics of tremor. To extract the tremor signal from the measured signal, a 4th-order Butterworth zero-phase band-pass
5.5 Results

Filter with cutoff frequencies of 2 and 30 Hz was used. Following signal filtering, the angular velocity of each target joint was obtained differentially from the two IMUs located on each side of the joint. Lastly, the angular displacement of each joint was integrated from the angular velocity using the Euler Forward method.

To quantify the performance of the WTSG suppressing tremor, the magnitude of the recorded tremor, root mean square (RMS) value, and the power of the tremor were calculated and are presented in this study.

5.5 Results

In this section, the performance of the proposed WTSG suppressing resting tremor, postural tremor, and kinetic tremor are presented. The results of resting tremor suppression are presented in Section A, while the results of postural tremor and kinetic tremor suppression are presented together in Section B.

5.5.1 Validation of the WTSG on Tremor Suppression: Resting Tremor

Fig. 5.11 shows the comparison between the unsuppressed tremor and the suppressed tremor in both time domain (Left column) and frequency domain (Right column) for all joints. The curves in blue represent the unsuppressed tremor (monitoring) and the curves in red represent the suppressed tremor (active suppression). The magnitude of tremor velocity is presented as the index for comparison in the time domain, and power spectral density (PSD) for comparison in the frequency domain. It is clearly shown in the time domain comparison that the suppressed tremor has a lower magnitude than the unsuppressed tremor for all joints. The overall RMS magnitudes of the unsuppressed tremor are 32.7 degrees per second (dps), 36.7 dps, and 27.2 dps for the index finger MCP joint, thumb MCP joint, and the wrist, respectively. The overall RMS magnitudes of the suppressed tremor are 8.8 dps, 7.1 dps, and 3.9 dps. Therefore, the overall tremor magnitude suppression percentages are 73.1%, 80.7%, and 85.5%. Similar results were obtained from the frequency domain analysis—each harmonic of the suppressed tremor has a lower PSD than the counterparts from the unsuppressed tremor. It is also shown in the results that the suppressed
5.5 Results

Figure 5.11: Comparison between the suppressed resting tremor (red) and the unsuppressed resting tremor (blue) from the index finger (IF), the thumb (T), and the wrist (W) in both time domain (left column) and frequency domain (right column). The resting tremor angular velocity was used in this comparison. The numbers in the figures in the right column show the frequency of the nearby harmonics.

The tremor has reduced frequency compared to the unsuppressed tremor. The specific frequency value of each harmonic is indicated in the right column of Fig. 5.11. The mean frequency reduction for the first harmonic across all three joints is 0.8 Hz, 1.8 Hz for the second harmonic, and 2.3 Hz for the third harmonic.

To evaluate the suppression of each tremor harmonic, the extracted tremor signal was further broken down into three separate harmonics. The time series of each harmonic is shown in Fig. 5.12. The left column shows the unsuppressed tremor harmonics from all joints, the middle column shows the suppressed tremor harmonics from all joints, and the right column presents the percentage RMS reduction in the magnitude of each tremor harmonic. The first harmonic is shown in blue, the second is shown in red, and the third is shown in green. It is visually apparent that each harmonic was suppressed significantly by the WTSG. The comparison of the RMS value for each harmonic
Figure 5.12: Comparison between the first three harmonics of the unsuppressed resting tremor (first column) and suppressed resting tremor (second column). Note that each y-axis has different scale. The RMS magnitude and the percentage suppression of each tremor harmonic are shown in the third column.

shows that the first harmonic has the highest mean suppression percentage (80.4%) across all joints, followed by the second (71.1%), and third harmonics (67%).

5.5.2 Validation of the WTSG on Tremor Suppression: Action Tremor

Similar to the analysis of the resting tremor suppression, the analysis of the postural tremor suppression was conducted in both the time domain and the frequency domain (Fig. 5.13). The figures in the left column show the magnitude of the unsuppressed tremor (blue curves) and the suppressed tremor (red curves) for all three joints. Visual inspection of both tremor signals clearly shows that the tremor motions present in the subject’s joints were suppressed significantly by the WTSG. The overall RMS magnitudes of the unsuppressed tremor are 27.4 dps, 22.6 dps, and 22.7 dps for the index finger MCP joint, thumb MCP joint, and wrist, respectively. The overall
5.5 Results

Figure 5.13: Comparison between the suppressed postural tremor (red) and the unsuppressed postural tremor (blue) from the index finger (IF), the thumb (T), and the wrist (W) in both time domain (left column) and frequency domain (right column). The postural tremor angular velocity was used in this comparison. The numbers in the figures in the right column show the frequency of the nearby harmonics. RMS magnitudes of the suppressed tremor are 8.2 dps, 4.6 dps, and 4.3 dps. Therefore, the overall tremor magnitude suppression percentages are 70.2%, 79.5%, and 81%. The corresponding frequency representation of each tremor signal is shown in the right column. The suppressed tremor presents a lower power across all harmonics compared to the unsuppressed tremor. Similar to the results from the analysis of the resting tremor, the frequency of the postural tremor was also reduced when the WTSG was active. The frequency of each harmonic is shown in Fig. 5.13. The mean frequency reduction for the first harmonic across all three joints is 0.7 Hz, 1.4 Hz for the second harmonic, and 2 Hz for the third harmonic.

To investigate the suppression performance in each tremor harmonic, the first three harmonics were extracted from the postural tremor signals. The time series of each harmonic is shown in
Figure 5.14: Comparison between the first three harmonics of the unsuppressed postural tremor (first column) and suppressed postural tremor (second column). Note that each y-axis has different scale. The RMS magnitude and the percentage suppression of each tremor harmonic are shown in the third column.

Fig. 5.14. The left column shows the unsuppressed tremor harmonics from all three joints, the middle column shows the suppressed tremor harmonics, and the right column provides the percentage RMS magnitude reduction. The first, second, and third harmonics are shown as blue, red, and green curves, respectively. Similar to the results from the analysis of the resting tremor suppression, the WTSG suppressed a large amount of the postural tremor. The obtained RMS magnitude reduction for each harmonic shows that the first harmonic has the highest mean suppression percentage (78.7%), followed by the second harmonic (68.5%), and the third harmonic (66.1%).

Lastly, the performance of the proposed WTSG when suppressing kinetic tremor is shown in Fig. 5.15. The figure presents the angular displacement of the index finger MCP joint (first row), thumb MCP joint (second row), and the wrist (third row). The black dashed lines separate each
Figure 5.15: The angular displacement of the user’s joints during the action task. The red curves represent the measured motion, the blue curves represent the extracted voluntary motion, and the green curves represent the extracted tremor motion.

task repetition. The measured motion signal is shown in red, the voluntary motion and tremor motion extracted from this measured motion are shown in blue and green, respectively. The overall RMS magnitudes of the unsuppressed tremor (left column) are 0.4°, 0.38°, and 0.36° for the index finger MCP joint, thumb MCP joint, and wrist, respectively. The overall RMS magnitudes of the suppressed tremor (right column) are 0.19°, 0.17°, and 0.14° for the three joints. Therefore, the overall tremor magnitude suppression percentage are 52.6%, 55.5%, and 59.8% for the joints.

5.5.3 User Experience

User experience feedback on comfort and function of the WTSG was obtained at the end of the trial. The goal of obtaining this feedback was to gain insights to improve the design for future clinical trials of the WTSG. The received feedback is listed in Table 5.2.
5.6 Discussion

This study has described the development and validation of a new wearable tremor suppression device for managing hand tremor. The intention was to introduce a hand tremor suppression glove as an alternative approach to the traditional tremor management approaches. The validation of the proposed WTSG on an individual parkinsonian tremor is novel insofar as no study has been conducted to suppress tremor in the fingers independently. Through the use of a cable-based transmission system and IMUs, the proposed WTSG features a compact and unobtrusive design. The successful experimental validation of the proposed WTSG on a human subject has demonstrated the feasibility of using an externally worn exoskeleton-type device for tremor management.

The comparison between the data of the suppressed tremor and the unsuppressed tremor demonstrates that the majority of the tremor presented in the index finger MCP joint, thumb MCP joint and the wrist were suppressed by the WTSG. This result is supported by both the time domain analysis and the frequency domain analysis of all types of tremor. The breakdown analysis of each tremor harmonic has shown that the suppression percentage of the tremor harmonics are similar in all three joints. This result is expected, as the proposed WTSG does not actively suppress tremor by actuating the target joints in the opposite direction, but instead considers tremor as a disturbance to the system. The analysis of the motion task indicates that the proposed WTSG can follow the voluntary motion of the subject while suppressing tremor. However, it also appears

Table 5.2: User Experience

<table>
<thead>
<tr>
<th>Comfort</th>
<th>Function</th>
</tr>
</thead>
<tbody>
<tr>
<td>* The glove is snug and comfortable</td>
<td>* Good ability to follow voluntary motion</td>
</tr>
<tr>
<td>* The use of velcro tape stabilizes the glove to the hand</td>
<td>* No noticeable overshoot from the motors</td>
</tr>
<tr>
<td>* No sweat or heat caused by wearing the glove</td>
<td>* No extra force from the motors on joints</td>
</tr>
<tr>
<td>* No noticeable physical discomfort or pain</td>
<td>* No extra force from the motors on joints</td>
</tr>
<tr>
<td>* The fit of the WTSG is good</td>
<td>during the action trial</td>
</tr>
<tr>
<td>* Cushioning is required for the interdigital folds due to the extra</td>
<td>* Does not make sudden movements</td>
</tr>
<tr>
<td>pressure applied by the index finger ring</td>
<td></td>
</tr>
</tbody>
</table>

5.6 Discussion

This study has described the development and validation of a new wearable tremor suppression device for managing hand tremor. The intention was to introduce a hand tremor suppression glove as an alternative approach to the traditional tremor management approaches. The validation of the proposed WTSG on an individual parkinsonian tremor is novel insofar as no study has been conducted to suppress tremor in the fingers independently. Through the use of a cable-based transmission system and IMUs, the proposed WTSG features a compact and unobtrusive design. The successful experimental validation of the proposed WTSG on a human subject has demonstrated the feasibility of using an externally worn exoskeleton-type device for tremor management.

The comparison between the data of the suppressed tremor and the unsuppressed tremor demonstrates that the majority of the tremor presented in the index finger MCP joint, thumb MCP joint and the wrist were suppressed by the WTSG. This result is supported by both the time domain analysis and the frequency domain analysis of all types of tremor. The breakdown analysis of each tremor harmonic has shown that the suppression percentage of the tremor harmonics are similar in all three joints. This result is expected, as the proposed WTSG does not actively suppress tremor by actuating the target joints in the opposite direction, but instead considers tremor as a disturbance to the system. The analysis of the motion task indicates that the proposed WTSG can follow the voluntary motion of the subject while suppressing tremor. However, it also appears
that the suppression level of the kinetic tremor is lower than the other two types of tremor (the suppression of resting tremor being at the highest). This is likely because the subject does not have severe kinetic tremor and the magnitude of the kinetic tremor is likely close to the actuation resolution of the proposed WTSG. A closer look at the left column of Fig. 5.15 shows that the magnitude of tremor varies in a motion cycle. An enlarged view of the index finger position profile is shown in Fig. 5.16. It is apparent that the magnitude of tremor is reduced when the joint position changes quickly. This sort of fast motion is indicated by a starting black solid dot and an ending green solid dot. Therefore, to improve the performance of the WTSG during the action task, in addition to improving the resolution of the actuation system, a tremor onset detection algorithm should be implemented.

The frequency analysis of the suppressed tremor has shown that the magnitude of each tremor harmonic was significantly reduced. Not only was the PSD of each harmonic reduced, but the main frequency of each harmonic was also decreased. It was found that the frequencies of the first, second, and third harmonics of the resting tremor were reduced by an average of 0.8 Hz, 1.8 Hz,
and 2.3 Hz in the three joints. Similarly, the mean frequency reduction of the first, second, and third harmonics of the postural tremor were 0.7 Hz, 1.4 Hz, and 2 Hz in the three joints. Similar trend was also observed in the studies that have investigated how weight loading affects tremor characteristics [44–46]. The cause of this frequency shift could be associated with the change of the natural frequency of tremor as a reaction to the mechanical stimulation through the afferent pathway. In addition, this change of frequency could also be an amplification of a non-tremor disturbance as a by-product of the use of a feedback control system.

Ideally, tremor motion could be completely suppressed if the WTSG was rigid, however, it is required that the WTSG have a minimum effect on the user’s voluntary motion. Therefore, the tremor and voluntary motion estimators play a significant role in determining the performance of the WTSG. In the proposed WTSG, although the use of a WFLC as tremor estimator has an advantage in real-time performance, the estimation accuracy is rather low compared to other complex estimators [9, 47–49], especially when the frequency of the voluntary motion is close to the frequency of the tremor. Therefore, the WFLC unfortunately contributes to the error of the voluntary motion estimation. As a result, the postural tremor suppression performance of the WTSG was compromised. The resting tremor suppression level is higher than the postural tremor suppression level since the estimated voluntary motion is almost zero, thereby resulting in a higher tremor magnitude reduction.

There are several limitations identified in this study. One of the limitations is that the proposed WTSG does not incorporate the prediction of voluntary motion initiation. Therefore, during the trial, when transitioning from a resting state to a moving state, the subject had to exert a higher-than-normal effort to compensate for the stall torque of the motors, since the control system cannot immediately recognize the initiation of the voluntary motion.

In addition, the analysis of the video from the experiment identified relative movement between the actuation blocks and the user’s arm. This is likely because the Velcro straps did not provide enough force to stabilize the WTSG onto the subject’s arm. This movement reduces the tremor suppression performance due to insufficient tension in the cables.

Lastly, it was noticed from the video that tremor in the proximal interphalangeal joints (PIP) and distal interphalangeal joints (DIP) remained active when the WTSG was activated. This effect
was studied in our previous work [38], showing that the magnitude of unsuppressed tremor in the
distal joint(s) increases when the tremor in proximal joint(s) is suppressed. Therefore, this impact
in the distal joints could be caused by the strapping of the forearm and the active suppression
in the wrist and MCP joints. In spite of the promising results obtained from the current trial,
the WTSG requires the integration of tremor suppression in the distal finger joints to have the
greatest impact on hand tremor.

5.7 Conclusion and Future Work

This paper has presented the development and validation of a WTSG. This device is able to
suppress tremor motion while allowing voluntary motion in the index finger MCP joint, thumb
MCP joint, and the wrist. The WTSG is equipped with five IMUs and three BLDC motors for
the measurement and control of the joint motion. A cable-based transmission system was used to
transmit force from the motors to the user’s joints. This design significantly reduces the size and
weight of the WTSG. An experimental evaluation of the WTSG was performed on an individual
with PD. The WTSG achieved an overall suppression of 73.1%, 80.7%, and 85.5% in resting tremor,
70.2%, 79.5%, and 81% in postural tremor, and 52.6%, 55.5%, and 59.8% in kinetic tremor in the
index finger MCP joint, the thumb MCP joint, and the wrist, respectively.

All three tasks have shown promising results for the use of the WTSG for suppressing hand
tremor; however, further refinement of the mechanical design and control system of the WTSG
should be undertaken in order to improve the overall suppression performance. Additional subjects
should be recruited in order to further validate the WTSG.

The limitations of the current WTSG have highlighted some very important aspects to be
investigated in future work. One is the inclusion of the tremor suppression in the PIP and DIP
joints. Since the suppression of tremor increases the magnitude of tremor in the unsuppressed
joints, the degree of tremor suppression afforded by the current version of the WTSG is insufficient
to perform activities of daily living (ADL). A study of individuals with PD performing ADL with
the WTSG should be conducted as the next step towards the transition of the WTSG from the
lab to the end user.
Bibliography


Chapter 6

Spatial Impact of Suppressed Tremor on Other Joints

This chapter is adapted from ‘Impact of Suppressed Tremor is suppression of proximal joints sufficient’, published in the Proceedings of the 2018 IEEE EMBS International Conference on Biomedical & Health Informatics, Las Vegas, NV, USA, March 4–7, 2018.

6.1 Abstract

Recent progress in wearable technology has made exoskeleton-type therapy devices a potentially viable alternative solution for parkinsonian tremor management. The target user group includes patients for whom current treatments have had no or minimal effect, or may cause major complications. So far, a number of tremor suppression devices have been developed. However, most of the studies and devices only considered the tremor in the wrist and elbow despite evidence that tremor is also present in the finger joints. The aim of this paper is to study the impact of suppressed tremor on the unrestricted joints, and to determine the importance of suppressing finger tremor in addition to the other most-studied joints. In this paper, tremor was analyzed using linear acceleration and electromyography (EMG) signals. The results show an increase in tremor magnitude in the unrestricted joints; however, EMG analysis did not show significant change in tremor muscle activity. This indicates that the increase in tremor motion may be the result of the
propagation of tremor from one joint to another.

6.2 Introduction

Tremor, being one of the most severe symptoms of Parkinson’s disease (PD), is often observed in body segments, such as the hands, arms, and legs. The presence of tremor significantly affects the daily life of patients, creating difficulty in many tasks such as eating and writing [1]. More importantly, it often results in social embarrassment that may lead to mental issues [2, 3]. Traditional treatments for tremor include medication and, in extreme cases, neurosurgery [4, 5]. These treatments are variably effective in reducing tremor, but they are also connected with a series of adverse effects and complications [6, 7]. Furthermore, there is a large proportion of patients (around 25%) for whom current treatments do not achieve a significant reduction of tremor [8].

Since Rosen’s experiment suggested that a mechanical loading orthosis could be used to selectively suppress tremor motion [9], a number of studies [10–17] have focused on developing tremor suppression devices. Elbow tremor suppression devices have been the main focus of some research groups [10–13]. Others have considered the importance of suppressing hand tremor; however, the majority of the research that focuses on hand tremor has concentrated on wrist tremor [14–17]. The suppression of finger tremor has not been considered.

In a previous study [18], our research group presented the biomechanical characteristics of hand tremor. It provides a reference for the design of tremor suppression devices for the hand, especially the fingers. However, it did not investigate the impact of suppressed tremor on the other joints. This information is important in determining which joints to suppress when designing a tremor suppression device.

The present study further investigates upper limb parkinsonian tremor with an emphasis on how the tremor from unrestricted joint(s) changes while the tremor in other joint(s) is suppressed. Specifically, the aim of this study is to quantify the change of tremor magnitude in the index finger (IF) metacarpophalangeal (MCP) joint, thumb (T) MCP joint, wrist (W), and elbow (E) to determine the importance of suppressing finger tremor in addition to wrist and elbow tremor, and to ascertain whether tremor from one joint propagates to other joints when suppressed.
6.3 Methods

6.3.1 Subjects

A total number of 14 subjects with PD participated in this study. All subjects were diagnosed and recruited by a neurologist. The arm that presented the greatest amount of tremor was selected for the trial. Based on the severity of the tremor of each subject, 12 subjects’ right hands and 2 subject’s left hands were recorded. All subjects signed a consent form to participate in the study. The experimental procedures were approved by the University of Western Ontario’s Health Research Ethics Board.

6.3.2 Experimental Procedure

Prior to the commencement of each trial, a customized motion-sensing system and surface EMG acquisition system were affixed over the patient’s hand, arm, and thorax. The motion-sensing system consists of seven 9-axis inertial measurement units (IMUs) (STEVAL-MKI108V2, STMicroelectronics®). The locations of the IMUs are labeled as red dots in Fig. 6.1. The distance between the IMU microchip and edge A of the PCB is 26 mm. Edge A of each IMU is placed as close as possible to the joint, hence IMUs on both sides have opposite orientations. The surface EMG sensors (Trigno™, Delsys, Inc.) were placed over the thumb thenar muscle group, finger flexors, finger extensors, wrist flexors, wrist extensors, biceps, and triceps muscles. The locations of the EMG sensors are labeled as green squares in Fig. 6.1.

After the placement of the sensors, each subject performed the following tasks:

1. Rest: the patient was asked to keep his/her arm in a rest position on a table for 60 seconds.

2. Outstretch: the patient was asked to maintain his/her arm outstretched for 60 seconds.

Three repetitions of each task were performed with the patient wearing a hand splint (Rolyan®, Kydex Orthosis), an elbow splint (Innovator®, Ossur), and no splint. The hand splint restricts the motion of the wrist and fingers, and the elbow splint restricts only the elbow motion. While the patient was performing the tasks, he/she was asked simple distracting questions (e.g., simple math questions) in order to distract their attention from their desire to suppress the tremor.
6.3 Methods

Figure 6.1: Sensor placement on an individual’s right arm. Solid red circles indicate the placement of the IMUs, and the placement of the EMG sensors is shown as green solid squares. A total of seven IMUs are placed from fingers to thorax, and seven EMG sensors are placed across the entire arm.

6.3.3 Data Recording and Processing

All data were collected and saved directly onto a personal computer (ASUS® desktop PC M51AC Series, Windows 8) using customized software. This software was designed to manage data transfer and storage from all of the sensors to the PC with a synchronized time stamp. The EMG data from the Trigno system were recorded using the EMGworks® software at a sampling frequency of 1926 Hz. The IMUs were interfaced with STC89C52RC microcontrollers using the I2C protocol. The data were sampled and sent to the PC at 100 Hz.

Data processing and analysis were performed offline using MATLAB software (Version R2013a, The Mathworks, Inc.). Calibration of the IMUs was conducted prior to data analysis. Following the calibration process, a gravity compensation algorithm was used to extract the linear acceleration. With the linear acceleration obtained, the tremor motion in each joint was calculated differentially from the two sensors located on either side of the joint. This data processing method is detailed in our previous study [18]. Prior to the data analysis, a 2nd-order Butterworth band-pass digital
6.4 Results and Discussion

A filter with cutoff frequencies of 3 and 30 Hz was used to eliminate the voluntary motion and high-frequency noise.

6.3.4 Statistical Analysis

The data were analyzed using a histogram to determine the distribution. Since the data appeared to be normally distributed, a repeated measures ANOVA was performed using the IBM Statistical Package for Social Science (SPSS, v.24). A $p$ value of less than 0.05 was considered to be statistically significant.

6.4 Results and Discussion

The trials were successfully conducted with no side effects or complications for the participants. Two separate analyses were completed using the motion data and EMG data. Both data sets were categorized into OFF and ON groups within each analysis. The OFF group consists of the data collected without the splint. The ON group consists of the data collected with splints, but only the joints and muscles that were not restricted by the splint were used for data analysis. The sensors used for the ON group are listed in Table I. The numbers correspond to the numbers shown in Fig. 6.1.

The quantification of tremor starts with the identification of tremor signals. The linear acceleration data set was used in the analysis of the joint tremor motion. A 1-second window swept through the entire signal to determine the start and the end of the tremor signal. The data in between were used to calculate the tremor RMS magnitude.

6.4.1 Paired Analysis of Joint Tremor Motion

Fig. 6.2 shows the RMS resting tremor magnitude from both the ON and OFF groups with each marker representing one joint of a single patient. The solid lines are the trend lines approximated with the patient data. The dashed line is the equilibrium line that separates the positive and negative change from the OFF to ON groups. The area above the equilibrium line indicates that the tremor magnitude from unrestricted joint(s) increases while the tremor in other joint(s) is
6.4 Results and Discussion

Figure 6.2: Distribution of the resting tremor RMS magnitude for both ON and OFF scenarios. Tremor in the index finger MCP joint, thumb MCP joint, wrist, and elbow are labeled as blue squares, red circles, purple triangles, and black stars, respectively. Solid lines represent the trend lines of each group. The equilibrium line is shown as the dashed line. The unit of each axis is gravity, g. The boxplot presents the unitless magnitude changes from OFF to ON, outliers are not shown for clarity.

suppressed. Conversely, the area below the equilibrium line indicates a decrease.

All patients show an increase of tremor magnitude in the index finger MCP joint and thumb MCP joint; two cases of decrease are found within the wrist and elbow data. The distribution of the markers shows that the tremor magnitude in the hand deviates from the equilibrium line more than the elbow. To quantify this deviation, the magnitude changes from OFF to ON are shown in the boxplot inset in Fig. 6.2. The mean ± SD of the magnitude change of the IF, T, W, and E are 0.54 ± 0.63, 0.96 ± 0.89, 1.24 ± 2.59, and 0.98 ± 2.09, respectively. Statistical analysis showed a significant difference ($p < 0.05$) between OFF and ON for all joints except the elbow ($p = 0.09$). This significant increase in tremor magnitude from OFF to ON shows the importance of suppressing tremor in distal joints, especially finger tremor, as finger tremor is ignored by most of the existing research.
6.4 Results and Discussion

Figure 6.3: Distribution of the postural tremor RMS magnitude for both ON and OFF scenarios. The unit of each axis is gravity, g. The boxplot presents the unitless magnitude changes from OFF to ON, outliers are not shown for clarity.

Fig. 6.3 shows the RMS postural tremor magnitude from both the ON and OFF groups. The magnitude changes from OFF to ON are shown in the boxplot. The mean ± SD of the magnitude change of the IF, T, W, and E are 0.50 ± 0.67, 1.02 ± 1.70, 0.49 ± 0.99, and 1.07 ± 3.14, respectively. This result is similar to the resting tremor. The statistical analysis showed a significant difference ($p < 0.05$) between OFF and ON for all joints except the elbow ($p = 0.18$).

These preliminary results indicate that the impact on unrestricted joints from restricted joints increases for both resting tremor and postural tremor.

6.4.2 Paired Analysis of Tremor EMG

Fig. 6.4 presents the EMG magnitude change during the trial. Each solid line represents the change of EMG magnitude from OFF to ON, while the boxplots on each side depict the distribution of the data from each group (left, OFF; right, ON). An average of 9 out of 14 of the samples showed an
6.4 Results and Discussion

Figure 6.4: EMG magnitude change from OFF to ON for all seven muscle groups. The resting tremor is shown at the top, the postural tremor at the bottom. The boxplots show the distribution of the patient EMG data. The y axis of each figure shows the value of the mantissa with an exponent of $10^{-4}$. The unit is volt.

There is an increase in tremor muscle activity for resting tremor and postural tremor. However, no significant difference was found within each muscle group for both types of tremor ($p > 0.05$).

The magnitude change of each muscle group is given in Fig. 6.5. Each boxplot represents the distribution of the magnitude change of the muscle group marked with a blue arrow. The resting tremor distribution is shown on the left and the postural tremor on the right. The highest averaged magnitude increase in resting tremor is achieved from the thumb thenar muscle group ($1.81 \pm 1.11$), followed by the index finger extensors ($0.70 \pm 1.28$), wrist extensors ($0.64 \pm 0.29$), triceps ($0.26 \pm 0.59$), biceps ($0.24 \pm 0.23$), wrist flexors ($0.11 \pm 0.26$), and index finger flexors ($0.03 \pm 0.123$). Although every muscle group has shown an increase in the average magnitude, no significant difference was obtained. As for the postural tremor, the highest increase is obtained from the thumb thenar muscle group ($1.29 \pm 0.95$), followed by the triceps ($0.63 \pm 2.39$), wrist extensors ($0.34 \pm 1.58$), index finger extensors ($0.32 \pm 0.66$), index finger flexors ($0.23 \pm 0.40$), biceps ($0.12 \pm 0.26$), and wrist flexors ($0.10 \pm 0.22$). Similarly, no significant difference was obtained.

The analysis of tremor EMG data has shown an increase in average magnitude from OFF to ON. However, no significant difference was obtained for any muscle group. This result indicates
6.5 Conclusion

The impact of mechanically-suppressed tremor on the tremor in other joints has been studied. A change in tremor motion magnitude was found in the unrestricted joints. This change varies from joint to joint for each patient, but it presents as a significant increase in all joints except the elbow across the entire patient population. No significant difference was obtained for the elbow, which may due to the low number of samples. The results from the hand joints indicate that it is not enough to suppress only the elbow and wrist tremor, but suppression of finger tremor should be considered as well. The analysis of muscle activity showed that the observed change was not

that mechanical suppression of the tremor motion is likely not affecting the generation of the tremor coming from the brain. Hence, the increased tremor motion in the unrestricted joint may be the result of the propagation of tremor from one joint to another.

Figure 6.5: Distribution of the EMG magnitude change for all seven muscle groups. The resting tremor is shown on the left, and the postural tremor on the right. Outliers are not shown for clarity.
significantly different. This may indicate that the increase in tremor motion is the result of the propagation of tremor from one joint to another.

As part of the future work, a spectrum analysis of the EMG signal will be conducted to investigate the possible mechanism of the change in tremor magnitude observed in the study. In addition, a wearable tremor suppression device will be developed with optimally selected actuators and a properly tuned control system using the quantitative results obtained from this study.
Bibliography


Chapter 7

Conclusions and Future Work

7.1 Concluding Remarks

Individuals with PD, who are receiving traditional treatments, often see improvement in symptoms such as bradykinesia and rigidity; however, there are currently no effective treatments for managing tremor. Given the side effects and complications of the traditional treatments, the emergence of wearable technology facilitates an externally worn tremor suppression device as a potential alternative approach for managing tremor. Through a literature review, it was identified that significant attention has been given to the elbow and wrist tremor by most researchers. Moreover, there are no commercially available active tremor suppression devices for patient use. Therefore, the main goal throughout the course of this thesis was to develop and validate a WTSG for people with parkinsonian tremor.

The work presented in this thesis established a fundamental step in developing and validating a WTSG for people with parkinsonian tremor. To assist with the development and validation of the WTSG, the tremors in the index finger MCP joint, the thumb MCP joint, and the wrist were characterized in both the time domain and the frequency domain, and an enhanced HWFLC-KF estimator was developed and validated with recorded tremor signals.

Both generations of the WTSG described herein share the same control system, sensing system, and electronic system. However, the cable-driven transmission system in the second-generation WTSG improved the efficiency and reduced the complexity of the entire actuation system. Prior to
the validation of the WTSG on a human, a physical tremor simulator was developed for evaluating the performance of the first-generation WTSG. The overall outcome of this validation proved the feasibility of using the WTSG to suppress tremor motion. This paved the way for a test on a human subject. To facilitate the testing on a human, a three-level safety measure was included in the development of the second-generation WTSG. The experiment showed promising results of the effectiveness of the WTSG; however, further optimization of both the hardware and software is needed to improve the suppression performance and user comfort.

7.2 Contributions

This work quantified parkinsonian hand tremor in both the time and frequency domains, developed an enhanced HWFLC-KF for separating the tremor signal from the user’s voluntary signal, and developed and validated a WTSG on a human subject. The specific contributions of this work are as follows:

- The kinematic and electro-physiological characteristics of parkinsonian hand tremor in both the time domain and the frequency domain were quantified. The joints of interest include the index finger MCP joint, the thumb MCP joint, and the wrist.

- The frequency analysis of the tremor data showed a distinct pattern that has been neglected by other researchers, which is that parkinsonian hand tremor consists of multiple harmonics. The frequency of the $k^{th}$ harmonic is roughly $k$ times the fundamental frequency.

- An enhanced HWFLC-KF was proposed and evaluated for the purpose of estimating tremor. The percentage estimation accuracy achieved from the proposed estimator was $96.3 \pm 1.7\%$, with average improvements of 28.5% and 48.9% over its lower-order counterpart and the WFLC.

- A physical tremor simulator was developed and evaluated for the validation of the first generation WTSG. This simulator can reproduce recorded tremor motion, tremor torque, and voluntary motion.
7.3 Limitations and Future Work

- A first-generation WTSG prototype was designed and validated on the physical tremor simulator using recorded tremor data from seven subjects with PD. This WTSG suppresses tremor in the index finger MCP joint, thumb MCP joint, and wrist in the direction of flexion and extension. Two assessments were performed to evaluate the performance of tremor suppression and following of voluntary motion. The tremor suppression assessment showed an overall tremor amplitude reduction of 85.0% ± 8.1%, and the following of voluntary motion assessment showed a RMSE of 14.2% ± 2.5% and a correlation coefficient of 0.97 ± 0.01.

- The mechanical design of the second-generation WTSG was improved to address the issues identified in the first generation WTSG.

- A pre-clinical trial was performed on one subject with parkinsonian tremor to validate the feasibility of using the second-generation WTSG suppressing tremor on humans. Results from the experiment showed an overall suppression of 73.1%, 80.7% and 85.5% in resting tremor, 70.2%, 79.5%, and 81% in postural tremor, and 52.6%, 55.5%, and 59.8% in kinetic tremor in the index finger MCP joint, the thumb MCP joint and the wrist, respectively.

- The effect of tremor suppression on the tremor in non-suppressed joints was evaluated. The magnitudes of the tremor motion in the unrestricted joints were increased when the tremor motion in the other joints were suppressed. This increase was particularly significant in the distal joints when the proximal joints were suppressed. This indicates that it is not enough to suppress only the elbow and wrist tremor, the suppression of finger tremor should be considered as well. The analysis of tremor EMG signal did not show a significant change, which indicates that the increase in the magnitude of tremor motion may be the result of the propagation of tremor from one joint to another.

7.3 Limitations and Future Work

Although significant progress in the development and validation of a WTSG was achieved, several limitations that need to be addressed have been identified:

1. It was identified during the validation experiment of the WTSG on a human subject that the
magnitude of the tremor in the middle finger, ring finger, and the pinkie remained unchanged. Moreover, the tremor in these fingers was not characterized.

2. The developed tremor estimator does not include tremor motion onset detection, nor does it include tremor motion prediction, and the parameters used in the estimator have not been optimized to the tremor of each individual.

3. The control system of the second-generation WTSG cannot react to the user’s motion in real time, therefore, there exists noticeable delay between the user’s motion and the reaction of the WTSG, which considerably influenced the performance of the WTSG.

4. The control system used in the current WTSG cannot adaptively optimize to different users and the types of tremor.

5. The validation of the WTSG on an individual with PD has shown promising results but issues with the mechanical design were also identified, such as the existence of cable slack, relative movement between the WTSG and the user’s hand, and the non-optimal placement of the glove components.

6. The validation of the current WTSG proved its ability to suppress tremor with minimal restriction on the user’s voluntary motion; however, the evaluation of the user performing ADL with the WTSG was not studied in this project.

7. The use of BLDC motors in the WTSG provided a certain degree of simplicity in the design process, and their power, bandwidth, accuracy, and speed make them feasible actuators for the WTSG. However, they do not have the inherent ability to replicate many key biological properties. This includes the ability to produce a large range of forces and strains, modulate compliance, produce linear actuation, and provide high power-to-weight ratios.

To address the aforementioned limitations, this work could be continued in several directions as discussed below.
7.3 Limitations and Future Work

7.3.1 Full Upper-Limb Tremor Suppression Device

Parkinsonian tremor in the index finger MCP joint, the thumb MCP joint, and the wrist were studied in this project for the purpose of developing a WTSG. It has been shown that a WTSG can suppress tremor in the index finger, the thumb, and the wrist, and this could help users perform some basic ADLs. Although the functionality of the index finger and the thumb are critically important to each ADL, the importance of the other three fingers should not be neglected. Therefore, the characteristics of the middle finger, the ring finger, and the pinkie could be quantified as the first step towards the suppression of tremor in these three fingers. In addition, a third-generation wearable tremor suppression device that includes suppression of all fingers and the elbow could be developed.

7.3.2 Tremor Prediction and Real-time Control System

The control system of the current-generation WTSG cannot suppress tremor without introducing delay. This drawback significantly affects the suppression performance. To improve the tremor suppression performance, it is possible to leverage machine learning methods, such as deep neural networks, to develop a dynamic tremor model that achieves better tremor suppression performance than is currently possible. This could be achieved by the following steps,

1. Analyzing tremor patterns using the EMG and motion data of the tremor in the target muscles and the joints to adapt to each user’s tremor behavior by developing the dynamic model of the WTSG.

2. Designing a real-time control system that predicts tremor motion ahead of time.

3. Incorporating tremor onset detection to improve the performance of the system in mitigating kinetic tremor.

7.3.3 Implementation of Smart Actuators

Considering the challenges of electrical motors replicating key biological properties, alternative smart actuators could be studied for implementation in the next generation WTSG, such as shape-memory alloy, pneumatic artificial muscles, electroactive polymers, and twisted coiled actuators.
Among these smart actuators, it was discovered by Haines et al. [1] that twisted coiled actuators produce all of the desirable properties of human muscle with higher output power than any other smart material. Therefore, the feasibility of integrating twisted coiled actuators into the next generation WTSG could be studied.

7.3.4 Incorporation of Multiple Suppression Modalities

The project presented in this thesis investigated the feasibility of suppressing hand tremor using a mechanical loading suppression technique. Other suppression modalities such as FES and mechanical vibration have also been investigated in the literature. Although each suppression modality has been shown to suppress tremor, they all possess drawbacks for patient use. During the course of this work, an interesting research question was discovered, i.e., what is the feasibility of adopting multiple suppression modalities in one WTSG? Following the idea of using multiple suppression modalities, it is possible to hypothesize that the drawbacks of each modality can be reduced and the suppression performance can be improved by using a smart control system that combines and adaptively adjusts the parameters of each suppression modality.

7.3.5 Functionality Test with WTSG

The goal of the WTSG is to suppress upper limb parkinsonian tremor, thereby enabling people with PD, who are disabled by the existence of tremor, to perform ADLs. An efficacy study that assesses the performance of the WTSG using both quantitative and qualitative analysis could be conducted. Such a study could determine how well subjects can perform tasks with and without the device. To quantify the ability of the user to perform ADLs with and without the device, the Action Research Arm Test could be considered [2].

7.3.6 Clinically Important Tremor Suppression Level

The development of the WTSG is not only an engineering project but also a clinical project. Without the clinical requirements and restrictions, an over-engineered but clinically useless robot could result. Studying the literature of tremor suppression devices, a very interesting common phenomenon was discovered, which is that researchers wish to design a perfect device that can
suppress tremor as close as possible to 100%. Such a phenomenon is very engineering-oriented. However, the level of suppression that is clinically important has not yet been studied. The quantification of a clinically relevant tremor suppression level could provide information on what level of tremor suppression allows people with tremor to perform ADL without limitations. Therefore, one important future work direction could be the investigation of the clinically relevant level of tremor suppression.
Bibliography


Appendix A

Permissions and Approvals

The following approvals are presented in this appendix:

1. Ethics approval for the PD hand tremor assessment from the Western University Health Science Research Ethics Board (HSREB)

2. Ethics approval for the validation of a wearable tremor suppression device on human subject from the Western University Health Science Research Ethics Board (HSREB)

3. Permissions for reuse of the published papers that constitute this thesis.
Western University Health Science Research Ethics Board
HSREB Annual Continuing Ethics Approval Notice

Date: January 05, 2017
Principal Investigator: Dr. Mary Jenkins
Department & Institution: Schulich School of Medicine and Dentistry/Clinical Neurological Sciences, London Health Sciences Centre

Review Type: Delegated
HSREB File Number: 106172
Study Title: Assessment of Hand Tremor

HSREB Renewal Due Date & HSREB Expiry Date:
Renewal Due - 2018/01/31
Expiry Date - 2018/02/20

The Western University Health Science Research Ethics Board (HSREB) has reviewed the Continuing Ethics Review (CER) Form and is re-issuing approval for the above noted study.

The Western University HSREB operates in compliance with the Tri-Council Policy Statement Ethical Conduct for Research Involving Humans (TCPS2), the International Conference on Harmonization of Technical Requirements for Registration of Pharmaceuticals for Human Use Guideline for Good Clinical Practice (ICH E6 R1), the Ontario Freedom of Information and Protection of Privacy Act (FIPPA, 1990), the Ontario Personal Health Information Protection Act (PHIPA, 2004), Part 4 of the Natural Health Product Regulations, Health Canada Medical Device Regulations and Part C, Division 5, of the Food and Drug Regulations of Health Canada.

Members of the HSREB who are named as Investigators in research studies do not participate in discussions related to, nor vote on such studies when they are presented to the REB.

The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

[Signature]
[Title]

[Signature]
[Title]

Ethics Officer: Erika Basile, Katelyn Harris, Nicole Kamki, Grace Kelly, Vikki Tran, Karen Gopaul

Western University, Research, Support Services Bldg., Rm. 5150
London, ON, Canada N6G 1C9 1-519-663-3036 1-519-850-2466 www.uwo.ca/research/ethics
Date: 27 May 2019  
To: AnaLisa Trejos  
Project ID: 110453

**Study Title:** Evaluation of a Hand Motion Tracker for Individuals with Parkinson’s Disease  
**Application Type:** Continuing Ethics Review (CER) Form  
**Review Type:** Delegated  
**REB Meeting Date:** 04/Jan/2019  
**Date Approval Issued:** 27/May/2019  
**REB Approval Expiry Date:** 04/Jan/2020

Dear AnaLisa Trejos,

The Western University Research Ethics Board has reviewed the application. This study, including all currently approved documents, has been re-approved until the expiry date noted above.

REB members involved in the research project do not participate in the review, discussion or decision.

Western University REB operates in compliance with, and is constituted in accordance with, the requirements of the TriCouncil Policy Statement: Ethical Conduct for Research Involving Humans (TCPS 2); the International Conference on Harmonisation Good Clinical Practice Consolidated Guideline (ICH GCP); Part C, Division 5 of the Food and Drug Regulations; Part 4 of the Natural Health Products Regulations; Part 5 of the Medical Devices Regulations and the provisions of the Ontario Personal Health Information Protection Act (PHIPA 2004) and its applicable regulations. The REB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

Please do not hesitate to contact us if you have any questions.

Sincerely,

Daniel Wyzynski, Research Ethics Coordinator, on behalf of Dr. Joseph Gilbert, HSREB Chair

*Note: This correspondence includes an electronic signature (validation and approval via an online system that is compliant with all regulations).*
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Appendix B

Curriculum Vitae

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**Post-secondary Education and Degrees:**

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2016–2019 Ph.D.
Biomedical Engineering

The University of Western Ontario
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Electrical and Computer Engineering

Tianjin University
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Control Science and Control Engineering
Tianjin Polytechnic University
Tianjin, China
2007–2010 B.E.
Automation

**Honours and Awards:** Western Graduate Research Scholarships (2016–2019)
Transdisciplinary Bone & Joint Training Award (2017–2019)
Western BME Travel Award (2019)
Western BME Travel Award (2018)

**Related Work**

**Teaching Assistant**

- *MSE 2201 – Introduction to Electrical Instrumention*
- *MSE 2202 – Introduction to Mechatronic Design*
- *ECE 2241 – Electrical Laboratory II*
- *ECE 3374 – Introduction to Electronics for Mechanical Engineers*
- *MME 4487 – Mechatronic System Design*

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**Publications:**


