**A Dynamic Approach to Balance Assessment using Self-Induced Perturbations**

by

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# ABSTRACT

A novel balance assessment method for older adults was developed using a balance platform that induces dynamic self-perturbations and which could use synchronization as an adjustable level of assistance. Rather than measure the subject's performance, the proposed balance assessment measures how much assistance they need to reach a standard level of performance. A swaying balance platform was constructed and instrumented to conduct the balance assessment with different levels of assistance. Nineteen healthy young adults were tested with self-perturbations introduced by following swaying visual instructions that changed frequency instantaneously. Experimental results were unable to confirm if synchronization was capable of providing assistance, nor whether the assessment outcomes could benefit from applying different levels of assistance. Future studies should focus on understanding how to ensure that synchronization occurs in the combined subject/instructions system.

# DEDICATION

À Danielle Suzanne Schutta Léger, mon amour éternelle.

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# 1 Introduction

Maintaining balance is an ability that most humans learn before the age of one. As explained by Winter et al. [1], and summarized by Bressel et al. [2], “Postural control or balance can be defined statically as the ability to maintain a base of support with minimal movement and dynamically as the ability to perform a task while maintaining a stable position.” The ability to maintain this balance allows people to perform everyday tasks and thus plays an important role in everyday life. However, there are several reasons why a person’s balance may decay. In addition to physical injury and general aging, many diseases such as stroke, Parkinson’s disease [3], Alzheimer’s disease [4], and osteoarthritis [5], can also lead to balance issues.

Research has suggested that the most common reason for falls in older adults is an inability to quickly adjust to self-imposed changes in their planned motions (a self-generated perturbation) [6], [7]. These motions are dynamic, meaning that their base of support moves over time. Although many existing balance assessment methods include dynamic aspects, they do not measure subject handling of dynamic self-generated perturbations. Also, though the literature shows that assessing balance with dynamic perturbations is more indicative of future falls, outcome measures produced when assessing older adults must overcome unsteady/noisy movements [8] (more erratic, less smooth) and a limited range of motion. Typical balance assessments are affected by these limiting factors and base their diagnosis solely on the performance measure of the test regardless of subject ability. Consequently, current dynamic balance assessments lose precision when

evaluating the likelihood of falls in older adults because they do not address noisy movements and a limited range of motion.

The use of an adaptive balance assessment approach that is able to respond to the abilities of the person being evaluated could boast several advantages. The most relevant benefit could be an ability to reduce the noise of the test introduced by confounding factors relative to underlying mechanisms of interest. This introduction discusses the impact of balance assessments, the implications of having older adults as a target population, why and how they fall, important elements in their balance assessment, the role of feedback, and the objectives of this research.

## 1.1 Impact of balance assessments

A person's balance relies on many different systems to function correctly. As inputs, balance relies on the eyes for vision, muscles and joints for proprioception, and the vestibular organs for head motion, position, and orientation. The weighting of these inputs may adapt in response to injury or disease, such as Alzheimer's, or simply with age [9]. Meanwhile, the mind must be capable of processing inputs to understand its current condition and required action accordingly. This, too, may be impacted by injury or diseases such as Parkinson's [3], Alzheimer's [4], multiple sclerosis [10], vestibular migraines, and ageing [11]. Finally, the reliability of muscles, bones, and their connective tissues plays a vital role in executing the commands given to them by the brain. Here, physical injury or diseases such as sarcopenia [12] and osteoarthritis [5] can lead to a degradation in the muscular and skeletal reliability of the body and increase the likelihood of an injury due to

accident or fall. As these three aspects of a person's balance become compromised, the risk of falls increases.

Falls can have consequences to a person's physical health, through an injury, but also to their mental health, such as a developed fear of falling [13], [14]. Either of these may lead to decreased physical activity, which often further exacerbates the problem. Moreover, falls account for almost 30% of injury related emergency room visits in Canada, resulting in the largest share of healthcare costs due to injury; an estimated $8.7 billion [15]. Due to the severity, frequency, and cost of falls in Canada and around the world, research in prevention and rehabilitation is of critical importance.

To avoid falls from occurring, effort must be focused on preventative methods. The first step in preventative methods are balance assessments. These assessments evaluate the state of a person's balance by challenging its inputs, information processing, and motor control. For subjects who are identified as being at risk of falling, preventative methods such as education, exercise, environmental modification, or the use of assistive aids can be implemented. When used appropriately, preventative methods help older adults avoid falls [16], [17], saving time spent in hospitals and money from lost income and/or surgical costs, and preserving quality of life [18].

## 1.2 Target population and mode of fall

As the leading cause of death, hospitalization, and permanent disability in Canada, the reduction of falls is critical to the sustainability of the healthcare system. Older adults (>65 years old) are especially at risk, as the rate of emergency room visits due to a fall increases with age [15] [19]. Moreover, the percentage of these visits leading to

hospitalizations can be two to five times higher compared to younger adults [15]. These statistics highlight the importance of understanding falls for this particularly vulnerable group.

A recent study by Robinovitch et al. [6], and another by López-Soto et al. [7], investigated how and why older adults fall in long-term care facilities and in hospitals respectively. In their study, Robinovitch et al. [6] captured video of 227 falls from 130 older adults. Of these, 41% of falls were found to be due to incorrect transfers or shifts of body weight, 21% from trips or stumbles, 11% from a hit or bump, 11% from loss of support with an external object, 11% from a collapse or loss of consciousness, 3% to slips, and a final 2% remained unknown for their cause. Similarly, López-Soto et al. [7] found that 32% of patients fell due to instability. These findings presented a shift in previous research; older adults are falling due to incorrect transfers in body weight more frequently than for any other cause of falls.

These weight shifts, as depicted in Figure 1.1 [6], occurred while subjects were attempting to quickly adjust due to self-imposed changes in their planned motions (a self-generated perturbation). These motions were also dynamic; their base of support moved over time. By including both dynamic and self-perturbation elements, a balance assessment could recreate the conditions in which falls typically occur for older adults, potentially increasing its predictive power.

## 1.3 Dynamic assessment

Based on the findings outlined in Section 1.2, a balance assessment for older adults should examine their responses under dynamic self-perturbations. Therefore, the subject's




**Figure 1.1. An example of an incorrect shift in weight that led to a fall.**

center of mass must move around during the assessment and subjects must react to perturbations initiated by their own actions (not an external 'push', for example).

Examining an older adults' response to dynamic self-perturbations, however, creates safety challenges given that dynamic self-perturbations, paradoxically, are a primary reason that they fall. Consequently, this work proposes an inherently stable balance platform design: without input, the mechanism returns the platform to a position that is level with the floor. Briefly, this balance platform uses an oscillatory swaying motion so that the subject can safely move their base of support while making the assessment dynamic. Instructions are given to the user to prompt the speed of these oscillations, and instantaneous changes in the prompted frequency are used to induce self-perturbations.

Older adults also have a limited dynamic range and produce noisier movements. The range in which balance may be safely assessed (a larger range than static conditions,

but not so big as to induce a fall) is therefore generally smaller for older adults [20]. Regardless of the reason(s) for this (whether natural decreases in strength or mobility, or due to injury or disease), the cumulative effect is a reduced dynamic range in which subjects may safely engage and in which balance may be assessed. Meanwhile, a dynamic environment with a greater range of movement can make it more challenging to distinguish between subjects at risk of falling and those who are not, by increasing within subject variability [8].

## 1.4 Feedback

In order to overcome the increased noise inherent in a dynamic balance assessment for older adults, this research proposes using feedback to assist the subject. Although the effects of feedback on balance training are documented [21], there is little research on the effects of feedback during balance assessments.

More specifically, this research proposes two distinct concepts of feedback. The first concept is within-trial feedback, which provides information about the instructions to the subject, and about the subject's movements to the balance platform. By doing so, the two subsystems may engage in synchronization: a control method in which two or more systems work to match each other more quickly than otherwise [22]. The feedback to the subject informs them of their position relative to the instructions and initiates self-perturbations by instantaneously changing the oscillation frequency of the instructions. The feedback to the system enables synchronization-based assistance by adjusting the instructions based on the user's performance.

For the second concept of feedback, rather than behaving like most balance assessments, which only use the test outcome measure to diagnose the subject, this research proposes that the finalized balance assessment uses an adaptive feedback method that iteratively adjusts the amount of synchronization-based assistance needed to reach a certain level of performance. The level of assistance required will then be used as the outcome metric, rather than the actual performance.

## 1.5 Objectives

This research contributes to the long-term objective of developing a clinically viable balance assessment tool that is automated, usable by older adults, and captures important diagnostic information about the subject's balance, through their ability to regulate dynamic self-generated perturbations. To fully evaluate the efficacy of this tool for clinical use, its ability to predict the likelihood of falls in older adults will need to be evaluated. The research and development process for this requires several steps, including the work presented here, future studies with target populations, and future longitudinal observational studies to determine the accuracy of the tool's fall prediction.

The objective of this study comprises the first step towards this long-term objective. As mentioned in Section 1, current clinically used balance assessments lack specificity and sensitivity, in part due to the noisy, less controlled movements of older adults. Unlike most balance assessments, which only use the test outcome measure to diagnose the subject, this research proposes that the balance assessment uses an adaptive feedback method, which iteratively tests subjects to find how much assistance they need to reach a standard level of performance.

Though this adaptive methodology could not be applied in this work, as the effect of synchronization on the outcome measure is the focus of the investigation in this thesis, the system behavior is explored at several different levels of synchronization. The collection of these behaviors is then used to understand how this adaptive methodology may affect the balance assessment when fully implemented. The objectives covered in this research include the development of an inherently stable balance platform prototype, the evaluation of a foundational methodology based on user assistance (as provided by synchronization), and the evaluation of adaptively changing user assistance based on trial performance.

It is hypothesized that the addition of adaptive synchronization, used to select appropriate levels of synchronization, reduces the noise and/or improves the testable range of a subject's balance assessment. This in turn allows for a more precise measurement of an older adult's ability to handle dynamic self-perturbed falls. This hypothesis is investigated along with the following specific aims:

- *Specific Aim 1:* Develop an inherently stable instrumented platform capable of assessing the balance response of older adults.

- *Specific Aim 2:* Design an assessment using the balance platform that evaluates the subject's ability to regulate dynamic self-perturbations.

- *Specific Aim 3:* Evaluate the promise of adaptive balance assessment by analyzing the effect of synchronization in several different states.

# 2 Justification & Feasibility

## 2.1 Testing strategies

The primary motivation for this research is the creation of a balance assessment for older adults, an exponentially growing demographic in Canada [23] and across the world. The current plethora of balance assessments fulfill various purposes but closer evaluation reveals that there remains room for improvement. In the pursuit of designing a more effective balance assessment, one starting point to consider is the reason why most falls occur in this target population.

Recent studies by Robinovitch et al. [6] and by López-Soto et al. [7] suggest that most falls by older adults in long-term care facilities and in hospitals are due to incorrect weight shifts. These weight shifts occur when residents attempt to adjust to a self-imposed change in their ambulation, such as when changing direction suddenly while walking. In other words, these falls occur due to self-generated perturbations during dynamic movements. The presence of these two elements, self-perturbations and dynamic movements, vary between balance assessments, as do equipment needs, staffing needs, clinical knowledge, and a host of other challenges. The proper use of these two cornerstones of balance assessment, however, should improve the assessment's reliability.

### 2.1.1 Perturbations

A balance assessment may use no perturbations, external perturbations, or self-perturbations in some or all of its tasks/components. In some cases [24], researchers also describe a fourth type of perturbation: sensory manipulation/perturbation. However, as it is not widely discussed, it is grouped with external perturbations for this review.

Several balance assessments have a few tasks that do not make use of perturbations, with only a few that make no use of perturbations at all [24]. These tasks do not ask or force the subject to change anything about their behavior or balance strategy throughout the task. Tests without any perturbations have received criticism as they do not sufficiently challenge subjects. Research points out that perturbations demand substantially more attention from a subject when maintaining balance [25]. Since older adults require more attention when maintaining balance [26], removing perturbations is likely to be a detriment to the test's sensitivity for this target population.

Tests with external perturbations attempt to recreate a scenario where a subject is thrown off their planned trajectory due to an external source. These sources include tripping/stumbling, being hit/bumped, losing support from a balance aid, and slipping. External perturbations are well suited to test a subject's reactive reflexes as the subject should not be able to predict the exact timing of the perturbation [24]. Research by Grassi et al. [27] has demonstrated that tests with external perturbations have a higher sensitivity in identifying balance impairments in minimally disabled multiple sclerosis (MS) patients, compared to tests without perturbations.

However, when using external perturbations, the risk of a fall during the task increases: a scenario best avoided when assessing the balance of older adults who tend to be more frail [28]. Though these falls can be contained in clinics using a harness, extra costs are required for the extra equipment and infrastructure. After a fall, bruises can still occur as well as injuries if the subject swings into other obstacles or if the harness is not fitted properly.

Finally, tests with self-perturbations pose a challenge to a subject's ability to adapt to their own planned trajectory, given their physical abilities. This type of perturbation comes from a subject's own actions and uses more anticipatory (a.k.a. proactive) reflexes [29]. Examples of self-perturbations in everyday activities include, but are not limited to, reaching movements, head turns while walking, sudden rises from a sitting position, and incorrect shifts in body weight as shown in the study by Robinovitch et al. [6].

The use of these three options for perturbations vary from test to test. A few balance assessments contain no perturbations at all, most contain some external perturbations, and many contain some form of self-perturbations. A summary of these assessments is shown in Table 1.

**Table 1. List of balance assessments organized by perturbations used and assessment setting.**

| | Clinical measures | Lab measures |
|---|---|---|
| Contains no perturbations | • Timed standing [24]<br>• steadiness in standing [24]<br>• the 10m walk [24] | • Short and long-term Lyapunov exponents [30]<br>• Maximum Floquet multipliers [30]<br>• Kinematic variability [30]<br>• Extrapolated center of mass (XCoM) [30] |
| Contains some external perturbations | • Tinetti POMA [31]<br>• BESTest [32]<br>• mini BESTest [33] | • Perturbed XCoM [30]<br>• Gait sensitivity norm [30]<br>• Foot placement estimator [30]<br>• Maximum perturbation [30] |
| Contains some self-perturbations | • Tinetti POMA [31]<br>• Berg balance test [34]<br>• step test [24]<br>• functional reach [24]<br>• reach test [24]<br>• step test [24]<br>• TUG [35]<br>• Dynamic Gait Index [36]<br>• Y balance test [37]<br>• Functional gait assessment [38]<br>• BESTest [32]<br>• mini BESTest [33]<br>• NeuroCom [39] [40]<br>• Biodex balance system SD [41] | |

### 2.1.2 Static and dynamic assessments

A review of the literature on falls has revealed that static and dynamic assessments each have unique advantages. Static tests are effective at evaluating the sensory receptors used for balance and past research has found that postural sway in static tests may predict falls [42] [43]. However, this claim has been contested for the case of assessing falls in the elderly, since static tests may only assess static balance and most falls for the elderly occur

during dynamic movements [25] [6] [7]. Dynamic tests, on the other hand, are more effective at evaluating gait, dynamic stability, and can include perturbations. Moreover, research has found that dynamic measures tend to be better at predicting falls [44].

Most clinical balance assessments actually contain both static and dynamic tasks. There are, however, some exceptions that contain only dynamic measurements. In laboratories, several dynamic measures are computed from gait characteristics to assess balance, whereas static measures include individual components of other balance assessments. A list of these assessments is shown in Table 2.

**Table 2. List of balance assessments organized by static or dynamic nature and assessment setting.**

| | Clinical measures | Lab measures |
|---|---|---|
| Static and dynamic elements | • Berg balance test [34]<br>• Tinetti POMA [31]<br>• Y balance test [37]<br>• Functional gait assessment [38]<br>• Dynamic Gait Index [36]<br>• BESTest [32]<br>• mini BESTest [33]<br>• Neurocom smart balance master [45]<br>• Balance system SD by Biodex [41] [46] | |
| Dynamic elements only | • TUG test [35] | • Short Lyapunov exponents [30]<br>• Long-term Lyapunov exponents [30]<br>• Maximum Floquet multipliers [30]<br>• Kinematic variability [30]<br>• Extrapolated center of mass [30] |
| Static elements only | | • Single leg stance [41]<br>• Static posturography [27] |

### 2.1.3 Balance assessment challenges

Aside from the choice of perturbations and the degree of movement in a test, there are several other factors that contribute to the success of a balance assessment. These considerations include price, subjectivity, ceiling and floor effects, sensitivity (the ability to identify all subjects at risk as such), specificity (the ability to identify all subjects not at risk as such), duration, portability, inter-rater reliability, repeatability (a.k.a. test-retest reliability), and more.

Importantly, pricing plays a substantial role in a clinic's decision to adopt a balance assessment. Unfortunately, many technology-based balance assessments, such as the Biodex System SD, the NeuroCom EquiTest, sequential weight-shifting [47], and other assessments based on the center of mass (CoM) or the extrapolated center of mass (XCoM) [30] using force platforms [48] [49] [50], require expensive equipment and maintenance. Bauermeister et al. [51] state that price is also one of the reasons why balance assessments do not directly measure each input, process, and output of balance.

Many of the less expensive balance tests contain subjective elements, which can lead to poor inter-rater reliability. Subjective measures often arise when balance assessment measures connect qualitative statements with a numerical scale. Balance related assessments such as the Berg balance test [52][53], Tinetti POMA [52] [53] [54], Activities-specific Balance Confidence Scale (ABC) [55], and the Timed up-and-go (TUG) [53] include such subjective measures.

Balance assessments may also exhibit floor or ceiling effects, where many subjects are grouped at the lowest or highest scores, respectively. Though many tests may be designed for a specific population, and thus a ceiling or floor could be explained if misused,

research has still found these issues when using the assessments as intended. Balasubramanian [56] found that the BBS, the Short physical performance battery (SPPB), the DGI, and the ABC all have ceiling affects when used for older adults. Meanwhile, Pardasaney et al. [57] found that the Tinetti POMA and the DGI exhibited a ceiling effect when testing community-dwelling older adults. Conversely, floor effects are documented for the SPPB [54] and in the Tinetti POMA when assessing balance in individuals with Huntington's disease [58] and in individuals in the later stages of Parkinson's disease [59].

The statistical levels of reliability (sensitivity and specificity) remain two of the most important performance measures of a balance assessment as they evaluate its ability to make correct claims. Several balance assessments have tunable thresholds that, if adjusted, may trade sensitivity for specificity or vice versa. A common practice when evaluating this threshold's optimum setting is to create a receiver operating characteristic (ROC) curve. Though some research has done exactly this [60] [61], most research conducted with these balance assessments typically employ a standard, default operating point value, limiting their effectiveness [25] [62].

Though some measures retain good reliability across performance evaluations in the literature, others, at times, receive much lower scores than originally reported (see Figure 2.1 [25], [54], [55], [58], [59], [62]–[75]). Tests that have a low reliability score may, however, still be used in clinics if they boast other advantages, such as being relatively simple, cheap, and quick to administer. One example of such a test includes the TUG test, which measures the time it takes a subject to stand up from a chair, walk three meters, turn, walk back to the chair, and sit down [35]. If the subject does not successfully complete the test under a given time threshold, which varies depending on the source, the subject may

be deemed at risk of falling. Although the TUG test is widely used due to its simplicity, it is limited by many factors, including the reaction time of the person timing the test. Research by Barry et al. [63] has actually concluded that the TUG test "should not be used in isolation to identify individuals at high risk of falls." Based on this conclusion, though the TUG test still has a purpose in clinics, a test with better reliability is still necessary to accurately identify patients in need of early intervention.

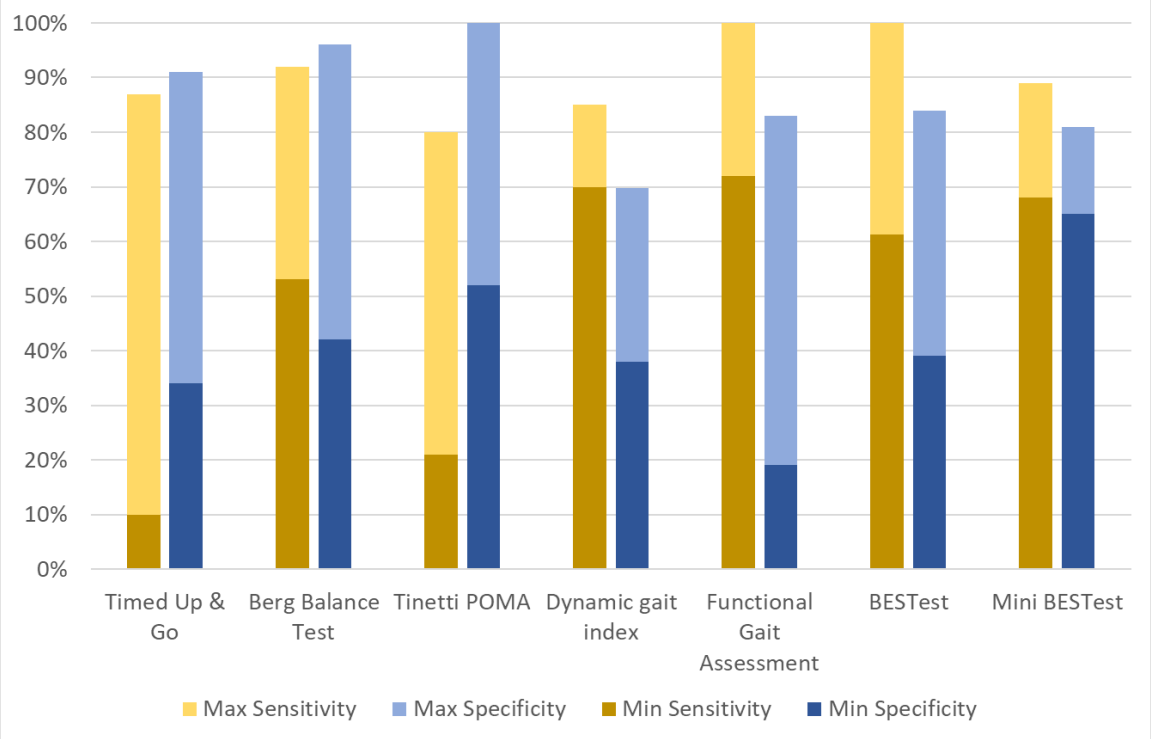



**Figure 2.1. Balance assessment reliability. This figure indicates the extremities of reported sensitivity and specificity values for several balance assessments typically used in clinics.**

### 2.1.4 Mechanisms

With the goal of minimizing the impact of these challenges, instrumented dynamic balance assessments hold some key advantages. Such assessments are objective, repeatable, and avoid issues of inter-rater reliability, since the subjects are evaluated by an

automated program on an instrumented device. Instrumented balance assessments also have the potential to be short in testing duration, and without ceiling or floor effects. However, depending on the design, the setup may be costly and immobile compared to non-instrumented balance assessments. For example, although the NeuroCom Balance Master (see Figure 2.2 [78],[79]) has good inter-rater reliability for most tasks [76], due to its size and complexity, it remains expensive and stationary. The Biodex balance system SD (see Figure 2.2) has also been found to have good repeatability [77], but suffers similarly in cost and mobility.



**Figure 2.2. The Biodex Balance System SD (left) and NeuroCom Balance Master (right).**

When designing a dynamic balance assessment, there are several mechanisms that may be used. The literature on balance theory and assessment frequently discusses the use of walking, treadmills, shifting platforms, and tilting platforms, to name a few.

Natural walking and treadmills boast the advantage of being very dynamic, because subjects will consistently raise their feet for locomotion. They aren't identical however, as recent research has pointed out that there are significant differences in gait variability and local stability between treadmills and natural walking [80]. Nevertheless, unconstrained

walking requires open space, which may be unavailable to some clinics and is difficult to monitor.

Shifting and tilting platforms may be contained within a smaller space, but do not typically facilitate reproducing gait. One notable difference between these two options is safety and stability. Though it may more closely resemble a slip-induced fall, a shifting platform as described by Oliveira et al. [81] would likely need to make use of external perturbations to remain dynamic. Doing so would require a harness, which, as mentioned in Section 2.1.1, can still lead to injuries. Meanwhile, a tilting platform, depending on its design, can remain dynamic and in a small space without the need of external perturbations [82].

One option for a tilting mechanism that may assess the balance of older adults is a wobble board (see Figure 2.3 [84]), as these devices are compact and typically use no external perturbations. For these reasons, previous work [83] at the Institute of Biomedical Engineering at the University of New Brunswick made use of a wobble board for balance assessment. However, wobble boards are more frequently used for balance training and their use in balance assessments has been limited [82]. This is likely due to their natural instability: wobble boards fall over when no input is given. This instability is not ideal for the balance assessment of older adults because it excludes subjects with insufficient balance control from taking the test and increases the risk of a fall.



**Figure 2.3. An example of a commercially available wobble board.**

### 2.1.5 Testing strategy for older adults

Given the surveyed information about perturbations, static and dynamic motion, general challenges, and mechanisms, the potential for improved methods of evaluating the balance of older adults gains definition. Self-perturbations (produced by the subject) represent the most common type of falls observed in older adults. A swaying motion can keep subjects dynamic while requiring little space. Instrumentation of the mechanism provides an objective and repeatable assessment as it eliminates the need for human raters. To date, these elements have not been combined for a balance assessment.

## 2.2 Feedback

Testing older adults dynamically poses a challenge for two reasons. First, older adults are typically limited in their range of motion as compared to younger adults. The range in which balance may be assessed, between no movement and movements that can cause a fall, is generally smaller for older adults due to physical disability or several age-related conditions such as Alzheimer's disease, Huntington's disease, Parkinson's disease, osteoarthritis, osteoporosis, and sarcopenia [20]. Second, humans naturally produce noisy signals with poor repeatability to activate their muscles [85] and age further increases this signal's variability [8]. The result is relatively high outcome variances over a small range, making it difficult to differentiate between subjects at risk of falling and those who are not. To remedy the issue of noise, this research proposes applying feedback.

Systems without feedback, also known as open-loop systems, are used in most balance assessments: the user completes a test (input) and is provided with a result (output), with little or no ability to respond. A block diagram of an open-loop system is shown in

Figure 2.4, where the test input is $r$, the input is modulated into necessary test parameter(s) as $u$ through operation in gain $k$, the test takes place in the plant $P$, and the test output is collected on the right hand side, as $y$. Note that triangle blocks provide a gain to their inputs and rectangular elements act as a sink, source or function, as determined by the context.

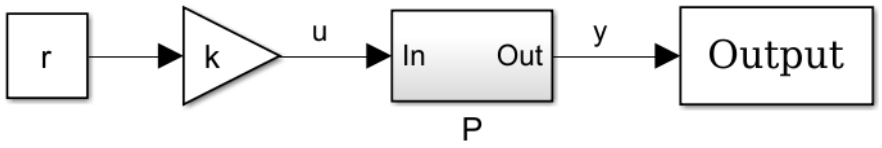



**Figure 2.4. Simplified open-loop control diagram.**

Because of the absence of feedback, the system's transfer function, the ratio of output over input, is modulated only by the gain $k$ (as shown Equation 1). Omitting feedback generally means the control system is simpler, cheaper to create and maintain, and is less likely to become unstable. However, disturbances and noise cannot be adjusted for automatically. The concept of an open-loop system is akin to, for example, a washing machine that cannot check the cleanliness of the clothes to determine when it is finished.

$$\frac{y}{r} = kP \tag{1}$$

Few current balance assessments leverage feedback, or a closed-loop system. In such systems, the subject can observe how well they are performing the task (output)

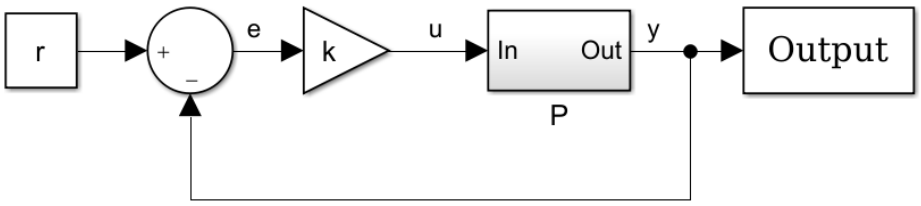



**Figure 2.5. Simplified closed-loop control diagram.**

relative to the instructions (input) and adjust their actions accordingly. A block diagram of

a closed-loop system is shown in Figure 2.5, using the same parameters as in Figure 2.4. Because of the addition of feedback, a denominator appears in the system transfer function, as shown in Equation 2.

$$\frac{y}{r} = \frac{kP}{1+kP} \tag{2}$$

This return of information allows the system to reduce the difference between the desired behavior of the system and the actual behavior of the system. This is comparable to cruise control on a car, where the engine adjusts its work output to maintain the instructed speed. Only instrumented balance assessments have feedback, such as the Neurocom smart balance master [45] and the Balance system SD by Biodex [46], as they require the return of information using sensors.

As discussed in Section 1.5, the long-term objective of this research is to develop an improved balance assessment for older adults. To do so, this thesis explores the use of feedback for two different applications: within-trial feedback to the user, and adaptive changing of difficulty across trials. The following subsections explain what these two forms of feedback are and why they may be beneficial to a balance assessment.

### 2.2.1 Within-trial feedback

As discussed in Section 2.1.5, a balance assessment strategy for older adults could make use of a swaying motion and have subjects follow instructions with self-perturbations. If designed like a gliding chair, a four-bar linkage can produce a swaying motion in a safe and stable tilting motion when subjects stand on the balance platform. However, the resulting small range of motion is an issue when providing visual instructions if they are to proportionally represent the mechanism on which the user would stand. A

simplified swaying animation, like a single line with a single pivot at its center, could be helpful to provide clearer instructions. These two mechanisms may be seen in Figure 2.6.



**Figure 2.6. The mechanism on the left is what is used in the proposed balance platform, while the mechanism on the right is a simplification used to provide instructions to the user.**

In dynamic and self-perturbed falls (see Section 2.1), subjects receive feedback from their surroundings through their vision, vestibular organs, and proprioception. A simplified swaying motion would alter the subject's understanding of how to interpret information from these sensors. Therefore, feedback could be applied within trials (while subjects are tested) by providing them with information about their angular position next to the simplified instructions for comparison. This would help subjects interpret their senses and better emulate the conditions in which most falls occur for older adults.

### 2.2.2 Synchronization

To improve the balance assessment's precision (the ability to differentiate between different subjects by their likelihood of falling from dynamic self-perturbations) synchronization may be applied. Observed in a wide variety of applied and social sciences, synchronization is a phenomenon in which 2 or more systems converge exponentially over time by adjusting based on the behavior of the other system(s) present. By applying synchronization, systems involved converge to the same state at an exponential rate [22].

The idea of using synchronization originated from a Dutch physicist named Christiaan Huygens as he observed the pendulums in two neighboring clocks, starting from

different initial conditions, converge to a synchronized state [86], [87]. Other examples include people walking side by side and eventually matching up their gait, an applauding audience, musical rhythm, or the bioluminescent patterns of a group of fireflies.

To demonstrate the concept of synchronization, take the example of Van der Pol oscillators as two-way coupled oscillators presented by Wang et al. [88]. If synchronization is applied to the two oscillators ($\mathbf{x}_1$ and $\mathbf{x}_2$), the states ($x$ and $\dot{x}$) would converge exponentially over time and eventually remain identical. A similar effect occurs at times where the oscillators become anti-synchronous, meaning that the states have identical magnitude but opposite signs. For the example of two Van der Pol oscillators, Equations 3 and 4 represent the sum of forces for both oscillators, where $\alpha$, $\omega$, $\kappa_1$, and $\kappa_2$ are system parameters.

$$\ddot{x}_1 + \alpha(x_1^2 - 1)\dot{x}_1 + \omega^2 x_1 = \alpha\kappa_1(\dot{x}_2 - \dot{x}_1) \quad (3)$$

$$\ddot{x}_2 + \alpha(x_2^2 - 1)\dot{x}_2 + \omega^2 x_2 = \alpha\kappa_2(\dot{x}_1 - \dot{x}_2) \quad (4)$$

The right-hand side of these equations represent the coupling function which works to reduce the difference between the two oscillators' states. By providing this feedback to each oscillator, they may be able to converge exponentially through synchronization. In this system, if $\kappa_1 + \kappa_2$ is greater than 1, then the two oscillators will synchronize as seen in Figure 2.7 [88].

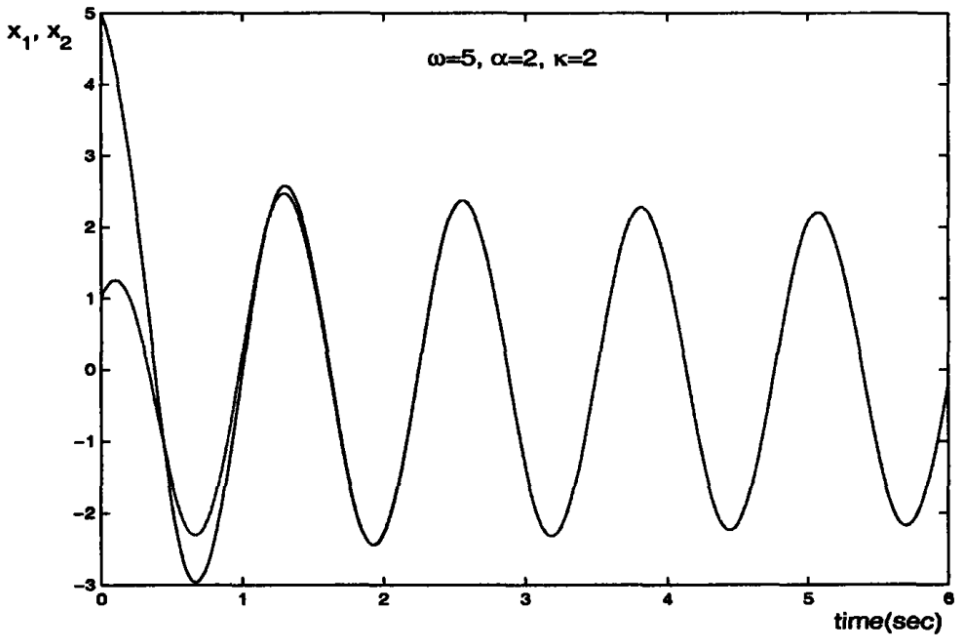



**Figure 2.7. Two Van der Pol oscillators synchronizing over time.**

### 2.2.3 Adaptive synchronization

Consider a balance assessment as represented by the control diagram presented in Figure 2.8. In this figure, $r$ represents the test's desired performance/outcome measure of the subject, $k$ is a gain which transforms this desired outcome to an input to the plant, $d$ represents any disturbances (lower frequency error which occurs before or during testing), the plant block $P$ represents the test which produces a performance/outcome measure, $n$ represents measurement noises (higher frequency error which occurs when the outcome measure is observed), and $y$ represents the outcome measure with noise. This simplified model represents a typical balance assessment since the outcome measure (corrupted by disturbance and measurement noise) is used by clinicians to evaluate the subject. In these open-loop balance assessments, the outcome measure is not used to adapt any future trials.

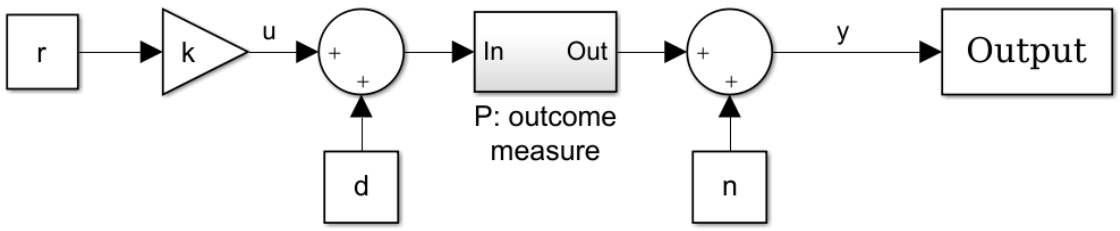



**Figure 2.8. The control diagram for an open-loop outcome measure balance assessment.**

Conversely, in the simplified model shown in Figure 2.9, trials run one after another until the desired outcome measure $r$ and the outcome measure $y$ are the same (within some tolerance). This is achieved by adjusting the gain produced in the block $k$, which is later used in the test of block $P$, between each trial. These adjustments are made proportionally to the error $e$ between the desired outcome measure $r$ and the outcome measure $y$. In this new suggested model, the outcome measure $y$ is no longer the output used by clinicians. Instead, the assistance provided $k$ needed to obtain the desired outcome is the output used by clinicians.

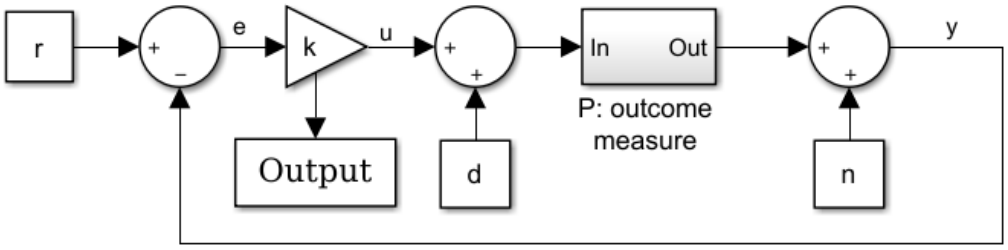



**Figure 2.9. The control diagram for a closed-loop outcome measure balance assessment.**

With respect to this thesis, the gain $k$ is the synchronization gain, which adjusts the degree to which the instructions are influenced by the subject's instantaneous error and represents how much assistance is provided during the test block $P$. The intention of synchronization is to assist subjects in achieving higher levels of performance and to

measure how much assistance is required to reach a preset level of performance. Note that this form of adaptive synchronization is not directly tested in this research as the effects of synchronization, used in the context of assistance, are unverified. Instead, several different values of synchronization gains are used in a random order (to reduce learning effects during analysis) and the results are analyzed. If successfully implemented, the feedback loop that enables adaptive synchronization offers some potential key advantages.

The first advantage of feedback is that the system becomes linearized. As shown by comparing the output $y$ over input $r$ ratio of the open-loop system (Equation 6) and the closed-loop system (Equation 7), adding feedback linearizes this transfer function if $|kP| \gg 1$. Moreover, if the output $y$ produced by the plant $P$ is calculated with nonlinear formulas, their effect on the transfer function is linearized since $P$ is present on both the numerator and the denominator. The benefit of a linear (or near-linear) transfer function is that the output of the system becomes easier to measure. In the case of this balance assessment, the output takes the form of the gains used for feedback, from the subject to the instructions, for a given trial.

$$Open-loop\ trasnfer\ function = \frac{y(s)}{r(s)} = kP \qquad (6)$$

$$Closed-loop\ trasnfer\ function = \frac{y(s)}{r(s)} = \frac{kP}{1+kP} \qquad (7)$$

As for the second benefit, the output should be more robust to perturbations since the comparison between the desired outcome and the outcome measure create a compensatory reaction. In the American Graduate Record Examination, for example, the desired outcome is the student's actual knowledge level and the outcome measure is the final score given by the test. Rather than giving a set series of questions, the test adaptively

determines the knowledge of the student by using an adaptive staircase. For a given amount of time, this procedure enables the test to reach a higher level of precision. If the student makes a mistake, the following question is easier. If the student is correct, the following question is harder. Despite any accidental disturbances introduced in the system, such as the student making a mistake uncommon for their knowledge level, the closed-loop test method narrows in on the outcome measure that matches the desired outcome.

# 3 Methods

As outlined, the long-term objective of this research is to develop a clinically viable balance assessment that is automated, usable by older adults, and captures important diagnostic information about the subject's balance via their ability to regulate dynamic self-generated perturbations. As a first step, a physical balance platform prototype was developed for use in this study. To achieve the long-term objective, the final design should ensure an automated and safe balance assessment. After prototype completion, software for testing and its protocol were developed. Finally, data was collected and analyzed using metrics based on channel capacity, to measure both subject performance and the effect of synchronization, and an assortment of other preliminary measures. This section details the methodology adopted to achieve these aims.

## 3.1 Hardware and instrumentation

Inspired by past research with wobble boards [83], the mechanics of this novel balance platform were designed to be stable: with no input, the platform remains in a horizontal position. This design decision was made to improve safety, as it was less likely to induce falls: an important consideration given the target population. To achieve this property, the balance platform uses a four-bar linkage mechanism very similar to the one used in glider chairs as shown in Figure 3.1.

This mechanism was chosen for the effect it produces as the user's center of gravity (CoG) moves in relation to the center of rotation (CoR) (identified by the intersection of the two long dotted lines in Figure 3.1). In general, the CoG tries to remain below the CoR because of gravity. As the platform moves to either side, the user's CoG moves in the same

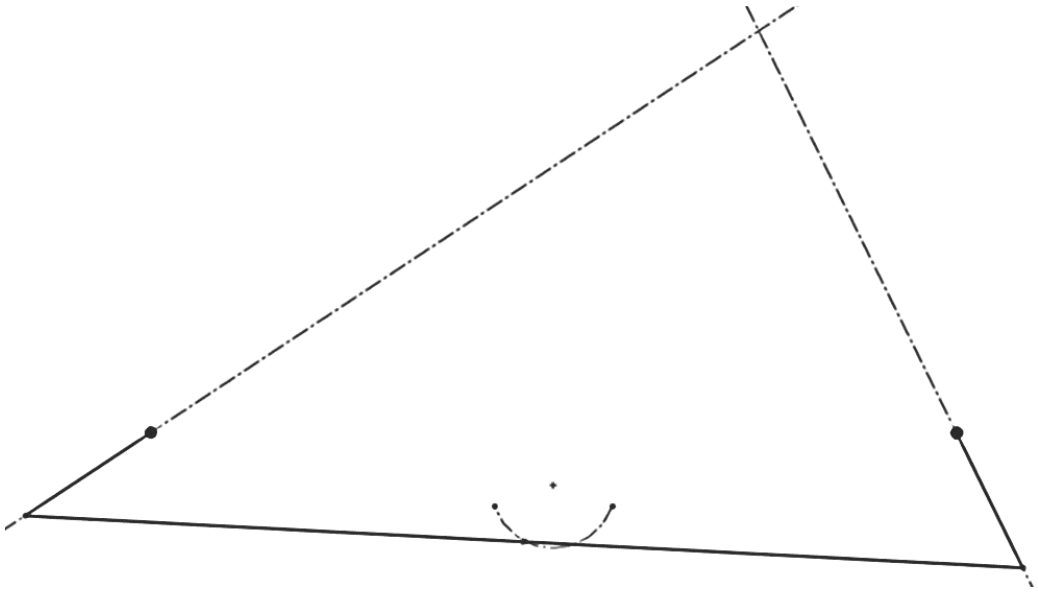


**Figure 3.1. Four bar linkage mechanism used by the balance platform. The two largest circles indicate fixed points of rotation. The small curved line at the bottom is a traced path of the platform's center. The long dotted lines, on each side, intersect above at the instantaneous CoR of the platform.**

horizontal direction as the CoR because of the four-bar linkage chosen. As a result, the platform actively works to keep the user from falling over. The torque applied by the horizontal offset of the CoG from the CoR was smaller for this design than for a pendulum-like platform. The prototype, built using this four-bar linkage design, is shown in Figure 3.2.

Four Hall-effect sensors were used to identify the angular position of the platform. Three Hall-effect sensors were placed in one corner pillar, and a fourth was placed in a corner diagonal to the first three (sensor locations are demarked with black circles in Figure 3.2). Magnets were placed inside the linkages, which allow the platform to swing as magnetic field sources for the Hall-effect sensors. This array of sensors allows for a full coverage of the platform trajectory, always keeping at least one sensor within a usable range. Each Hall-effect sensor was low-pass filtered using a $0.1\ \mu F$ capacitor (rated for

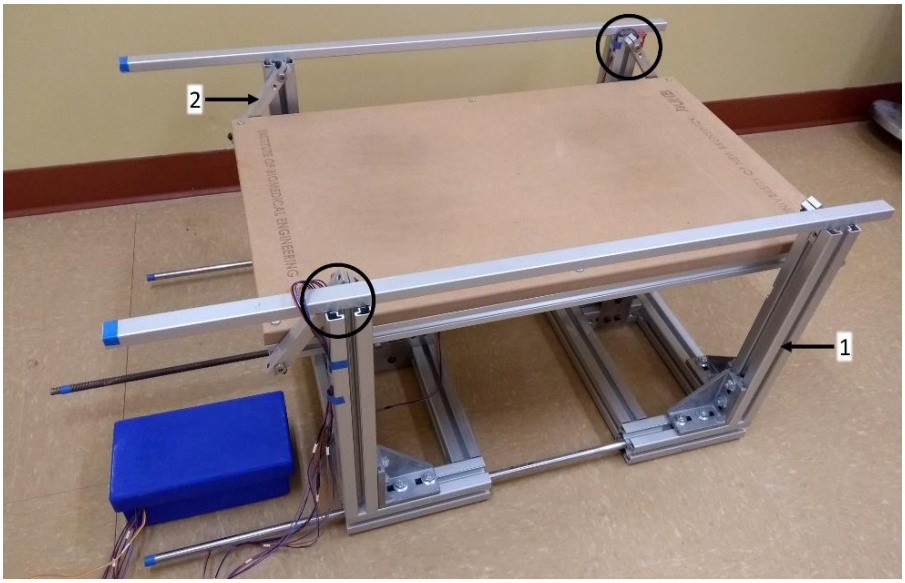



**Figure 3.2. The balance platform prototype with electronics. An example of a "corner pillar" is indicated by label 1, and label 2 for a "linkage". Black circles indicate the location of the hall-effect sensors which were connected to a microprocessor in the blue box on the bottom left corner.**

±20% and 50V) to reduce noise in the signal. Finally, an nRF52832 breakout board by Sparkfun was used to sample the analog voltage signals with a 14 bit resolution every 10ms and forward them to a host PC.

## 3.2 Data processing and visualization

Arduino's integrated development environment (IDE) version 1.8.5 was used to read signals from the Hall-effect sensors. These sensors produced a voltage that was proportional to the local magnetic field strength. As strong magnets were placed in the platform linkages, the voltage readings produced by the sensors were used to calculate the platform angle by interpolation with a calibration data set as shown in Figure 3.3. Calibration data were gathered regularly prior to experiments to ensure that any changes in the platform or sensors were captured.

With four Hall-effect sensors, four interpolated angles were available when testing. These angles were combined by weighting each angle inversely proportional to the magnitude of the voltage it originated from, as the sensors were more sensitive to movement at lower voltage readings. As sensors surpassed 2400mV, the weighting would be set to 0, eliminating the sensor's contribution to the angle computation. Similarly, weighting would eliminate sensor information when it neared a trough (see black vertical lines in Figure 3.3) as this range yielded higher uncertainty on either side of the trough.

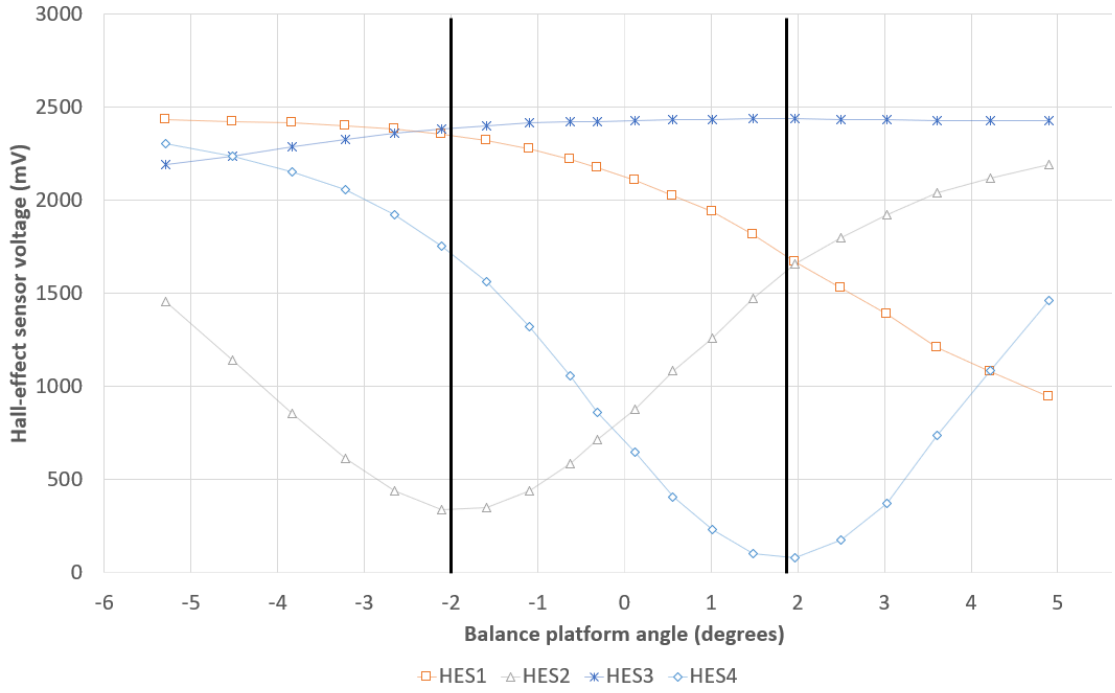



**Figure 3.3. Example Hall-effect sensor calibration interpolation chart used to determine angle based on voltage readout. The black vertical lines indicate two of the sensors' minima.**

The data were then imported to Matlab 2017a for further processing. A Kalman filter was used to the smooth the position estimates using angular position and velocity (which was derived from the position time series) as states. Measurement error was derived from an empirically determined confidence profile for each Hall-effect sensor based on its

inferred position. If a magnet was too far on either side of the sensor, the observed change in angle was quite noisy, whereas, if the magnet was too close, it was difficult to tell which side of the sensor it was on.

A graphical user interface (GUI), designed in Matlab, was used to display experiment settings (subject ID, perturbation switch and trial number) and a simple visualization for testing. This visualization consisted of two lines rotating about their center: one representing the current balance platform angle (green) and another the instructions given to the user to match (black) (see Figure 3.4). By animating both the instructions and the user as a comparison, the balance assessment provided feedback to the user throughout each trial.




**Figure 3.4. The GUI used during experimentation. The black line represents the instructions, while the green line represents the balance platform. The options on the bottom include (left to right) subject number, perturbation type (on or off), and trial number.**

## 3.3 Synchronization

This thesis proposed the use of feedback for two different applications in a balance assessment: within-trials and, in the finalized balance assessment, adaptively across trials. Within-trial feedback was applied using the GUI to better emulate dynamic and self-perturbed fall scenarios and to open the door to adaptive synchronization. The purpose of applying synchronization adaptively lies in the methodology planned for the finalized balance assessment: rather than measure a subject's performance, this balance assessment measures how much assistance a subject needs to reach a certain level of performance.

Though this research does not directly apply adaptive synchronization, as its ability to assist subjects had not yet been verified, it does offer a preliminary investigation of what would occur if it were applied by exploring several different levels of synchronization. Referring back to the Van der Pol oscillator example given in Section 2.2.2, this would be equivalent to testing different values for $\kappa_1$ and $\kappa_2$. It is expected that this should improve the diagnostic potential of the balance assessment by helping the test pull out the desired information with greater precision through the benefits of adaptive synchronization (discussed in Section 2.2.3).

Synchronization was implemented using feedback from the subject's angular error $\Delta\theta$ to modify the instructions given to the user in the GUI via the black line. The torque $T$ applied to modify the instructions was first calculated using Equation 8. In this equation, linear acceleration $a$ is multiplied by the moment arm $l$ and a mass $m$ (roughly equal to the mass of the subject and the moving part of the balance platform). The modification to the torque came from the angular error and a synchronization gain $k$ which represents the level of assistance provided. A synchronization gain of 0 implies no

assistance was given, whereas larger gain values provide more assistance. The new torque was then used to determine the angular position for the next time step.

$$\sum T = ml^2 \frac{d^2\theta}{dt^2} + k\Delta\theta = mla + k\Delta\theta \quad (8)$$

## 3.4 Data collection

To test the viability of the developed system, nineteen young healthy adults (10 males and 9 females), aged between 19 and 28 years of age (mean ± SD = 23.74 ± 2.68) were recruited. Subjects were required to pass a cognitive assessment test (to ensure they were mentally safe to complete a balance assessment) and have no injuries or disease that could affect their balance. Moreover, subjects had to be between 19 and 30 years of age so that future comparative studies could compare these data to that of older adults. All subjects gave informed consent as approved by the University of New Brunswick's research ethics board under REB# 2019-017.

### 3.4.1 Preparation

After obtaining informed consent, the experiment started with some preliminary tests. These tests included the Montreal Cognitive Assessment (MoCA), the Mini Balance Evaluation Systems Test (Mini BESTest), and a questionnaire. The MoCA's purpose was to screen the cognitive health of subjects; any subjects who failed the MoCA would have been excluded from the study for their safety. The Mini BESTest collects balance data on the subjects for future comparison as a well-documented balance test. The questionnaire gathered general information that could relate to the subject's balance. These included height, weight, age, sex, hand dominance, leg dominance, and balance affecting injuries or

diseases. Both the Mini BESTest and the questionnaire were collected for post-hoc analysis pending successful primary analysis.

Following preliminary testing, subjects were harnessed for their safety when using the balance platform. Subjects donned the harness, which was attached to a fixture in the ceiling with a slack rope to ensure minimum interference while testing, prior to stepping onto the balance board from a raised platform (see Figure 3.5). To use the device, subjects were instructed to sway gently left and right as instructed by a GUI on the monitor placed approximately 1 meter in front of the subject. Because of the design of the physical platform, swaying required simply leaning their weight onto their left or right foot.

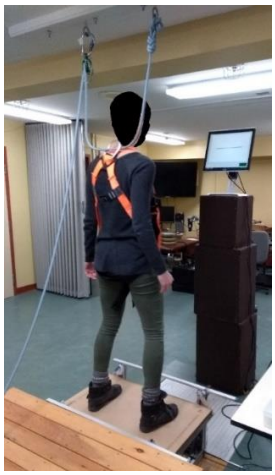


**Figure 3.5. A subject being tested on the balance platform setup.**

The GUI used to provide the prompt and feedback (as shown in Figure 3.4) included a green line, representing the balance platform's angle with respect to the ground, and a black line that they were instructed to match. The goal was, therefore, for the subject to orient the green line to be in line with the black prompted line as it oscillated back and forth. During the experiment, the black line oscillated at a nominal rate of 0.3 Hz, 0.4 Hz,

0.5 Hz, or 0.6 Hz. In some cases, the actual oscillation frequency varied based on the level of synchronization, as outlined in Section 3.3.

To reduce learning effects, each subject performed a familiarization session on the balance platform before testing began. This session was composed of 30 oscillations at a rate of 0.4 Hz, without any synchronization. This habituation also provided an opportunity for study requirements and clarifications before testing.

### 3.4.2 Testing

The testing protocol was structured to evaluate the combinations of two parameters: synchronization and frequency combination. The synchronization gain was varied using values between 0 and 5, in increments of 1, which modified the instructions based on the angular position error and angular velocity error. As shown in Section 3.3, a higher synchronization gain provided greater assistance as it created larger torques on the black instruction line.

Self-induced perturbations were introduced by changing the prompted frequency after the subject was given time to habituate to the previous frequency. All combinations of "from" and "to" frequencies were evaluated within each level of synchronization to ensure that all transitions could be evaluated. The only exception was the transition between the fastest and slowest frequencies (0.3 Hz to 0.6 Hz and 0.6 Hz to 0.3 Hz) as the jump between the two was found to be too large during piloting. This yielded ten frequency combinations, as shown in Table 3. During testing, the program prompted a given frequency for between 11 and 15 complete periods before instantaneously (and randomly)

switching to the next frequency. The order in which frequencies were tested was pseudo-randomized, but each frequency combination was tested.

**Table 3. Frequency combinations used.**

| | From | | | | |
|---|---|---|---|---|---|
| To | | 0.3 Hz | 0.4 Hz | 0.5 Hz | 0.6 Hz |
| | 0.3 Hz | N/A | √ | √ | × |
| | 0.4 Hz | √ | N/A | √ | √ |
| | 0.5 Hz | √ | √ | N/A | √ |
| | 0.6 Hz | × | √ | √ | N/A |

During piloting, the full suite of 60 test permutations (10 frequency combinations each for 6 synchronization gain values) was tested in 6 non-stop trials; one for each synchronization gain. However, due to mild fatigue experienced by subjects during pilot testing, each of these 6 trials were ultimately divided in two. The final protocol therefore consisted of 12 trials, each lasting 2 to 3 minutes, with an optional short break between each trial and a mandatory break after every 3$^{rd}$ trial.

## 3.5 Data analysis

The following sections present an overview of several metrics, calculated using Matlab 2017a, that were used to observe the direct effect of frequency and synchronization. Note that each subject had their trial data subdivided into sub-trials that analyzed each unique combination of synchronization gain and frequency transition (from one frequency to another). Following a set of preliminary analyses, outlined in 3.5.1, it was determined that two primary metrics of interest were required: a metric for subject performance and a metric to evaluate the effect of synchronization on the balance assessment's precision. Both metrics necessarily had to deal with changing bandwidths and noise levels. In order to

facilitate a comparison of these metrics across these changing levels, the concept of channel capacity was adopted in 3.5.2. The subsequent sections explain channel capacity, and why and how it was used to evaluate subject performance and to observe the effect of synchronization.

### 3.5.1 Preliminary performance measures

As a preliminary step in data analysis, the direct effect of frequency and synchronization on the other measures was observed using boxplots. These boxplots, developed in R 3.6.0, plotted several measures against synchronization while holding frequency or frequency combination constant, against frequency while holding synchronization constant, and against frequency combination while holding both frequency and synchronization constant. This step provided a first look at the data to confirm what was expected and to note any unanticipated effects. A series of performance measures, which included threshold-based measures and fit-based measures, were evaluated for appropriateness.

#### 3.5.1.1 Threshold-based measures

Four of the preliminary measures sought to identify how well the subject performed based on the time they took to meet one of two threshold conditions. Each of these times were normalized by the duration of all periods for the given frequency. The threshold conditions consisted of going below or staying below a given error threshold. This error took the form of either instantaneous angular error, windowed angular error, or subject-instruction delay. Subsequently, two additional measures were also calculated: the mean and median angular error after the windowed angular error threshold was passed.

Instantaneous angular error was calculated by finding the instantaneous difference in angular position between the subject and the instructions. For this measure, only the time at which the subject stayed below the error threshold was used, as pilot tests demonstrated that the instantaneous angular error would frequently venture near zero and quickly leave the threshold. As a logical next step, the windowed angular error used a sliding window that averaged angular error over the span of a single period. For this error measure, both thresholds types were used: time at which the subject went below the windowed angular error threshold and the time at which the subject stayed below the windowed angular error threshold.

Meanwhile, the delay error measure was calculated as the cross-correlation over a one period window of the subject's and instructions' angular positions. The sample window was then moved along the data to calculate how delay evolved over time. For this error measure, both threshold measures were used.

Finally, the mean and median error were based on the time at which the windowed angular error went below the threshold. So, the mean error was the mean windowed angular error after the threshold. Similarly, the median error used the same approach, but using the median error.

#### 3.5.1.2 Fit-based measures

The following preliminary measures were used to evaluate the SNR of both the subject and the instructions. Using the Fourier transform of the subject's or instructions' angular position, the signal was represented by the power at the target frequency, and the noise was represented by the power at all other frequencies. So, the measure would

demonstrate how closely the subject or the instructions (the instructions would stray from the target frequency due to synchronization) stayed at the target frequency. Using this method, 0 indicated that the subject or instructions never matched the target frequency and 1 indicated that they stayed perfectly at the target frequency.

In total, four SNR measures were implemented: two different methods were applied to each of the subject and the instructions. The mean SNR method divided the power of the target frequency by the sum of the power of the other frequencies (see Figure 3.6). The area SNR made use of the power spectrum versus frequency plot again by dividing the area under the target frequency range (one data point to the left and right of the target frequency marked in green on Figure 3.6) by the remaining area under the curve.

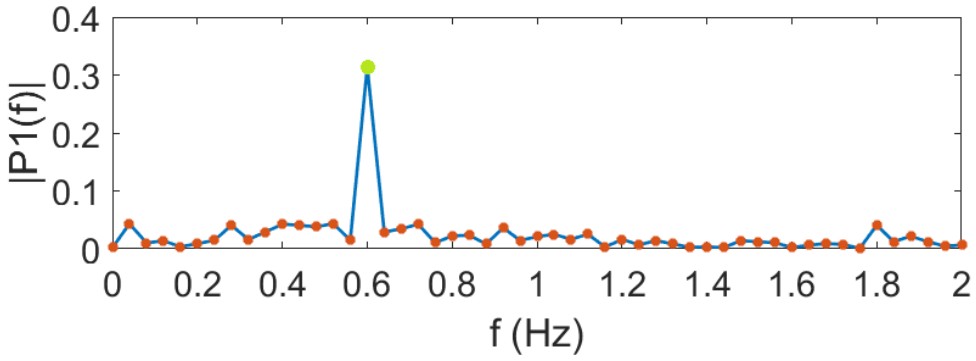



**Figure 3.6. Power of the Fast Fourier transform of a subject's angular position with respect to frequency. The target frequency is marked by a green dot and the noise data points by red dots. Note that the real frequency range analyzed went from 0 Hz to 10 Hz.**

The final additional measure was the coefficient of determination, $R^2$, between the subject's and the instructions' angular position over one sub-trial. This measure indicated how well the subject copied the target frequency of the instructions.

### 3.5.2 Channel capacity

Channel capacity, the maximum information rate that can be transmitted through a given channel, is a measure stemming from communications theory by Claude E. Shannon in 1948. The calculation of channel capacity can be performed using Equation 9, where $C$ is the channel capacity (bits/s), $B$ is the bandwidth of the channel (Hz), and $S/N$ is the signal to noise ratio (SNR). This Equation captures a tradeoff between bandwidth and noise.

$$C = B \log_2 \left(1 + \frac{S}{N}\right) \tag{9}$$

The application of channel capacity to human movement theorizes that a person has a specific maximum information rate that can be conveyed through a given channel. For example, a person can only transmit so much information when using a keyboard. If the person types faster (higher bandwidth) they become more prone to making mistakes (lower SNR). The opposite tradeoff could also occur if they were to type slower. The only way the person's channel capacity while typing would change was if they fundamentally changed their skill level through training.

### 3.5.3 Channel capacity for subject performance (CCP)

Subject performance was initially evaluated using a signal-to-noise ratio (SNR), which held the instructions as the signal and the subject's response as the signal with added noise. To calculate a dataset's SNR for this metric, the sample correlation coefficient "r" was assessed as a sliding window of one target period in length, averaged at each position, and mapped from a range of -1 to +1 to a range of 0 to infinity, as derived by Bershad and Rockmore [89], to emulate typical SNR measures.

Though synchronization's purpose was to help reduce the noise element of this SNR, it also created a tradeoff between this SNR and the bandwidth (regulated by the target frequency of the instructions). This resulted in challenges when extracting the true signal at any given point. Therefore, a new metric was needed that could measure subject performance without being affected as the subject changed speeds. Consequently, channel capacity was adopted as the subject performance/outcome measure (CCP).

This type of channel capacity measures a subject's performance at a synchronization value by averaging their performance at all frequencies tested for that synchronization value. CCP was calculated using the initial performance measure SNR (made using a mapped correlation coefficient [89]) and compensates for the tradeoff with the target frequency by using it as the bandwidth term "$B$" in Equation 9. This measure of performance has two applications: to help determine the effect of synchronization and, for future studies, to act as the performance measure which the user must reach with assistance from adaptive synchronization, as explained in Section 2.2.3.

Examining the relationship between CCP data and synchronization gains was necessary to confirm that an increase in the synchronization gain indeed assists the subject in achieving higher CCP values. Without this confirmation, it cannot be assumed that adaptive synchronization would function as intended.

### 3.5.4 Channel capacity for the effect of synchronization (CCS)

As mentioned in Sections 2.2.2 and 2.2.3, the use of synchronization may enable the implementation of adaptive synchronization in a finalized balance assessment, as it could adaptively vary the level of assistance to the subject. This level of assistance would

be adjusted adaptively and would represent aid to ensure that the task difficulty was appropriate for the user.

The main reason for using adaptive synchronization was the anticipated benefit of increasing the precision in subject performance scores, CCP, by reducing its noise relative to its bandwidth. Doing so would help clinicians better differentiate between subjects who are at risk of falling from those who are not. Part of the 3rd specific aim for this study was to evaluate this anticipated benefit.

Because of the tradeoff between bandwidth and noise noted in Section 3.5.2, channel capacity was also used to measure the effect of synchronization (CCS) on the repeatability/precision of the performance measure by accounting for exchanges between bandwidth and noise. A CCS value quantifies the repeatability/precision of CCP for all subjects, at all frequencies tested, at a specific synchronization gain value.

To calculate CCS, Equation 10 was adapted from Equation 9. In this formula, the bandwidth was replaced by the standard error of CCP, where each CCP value represents a subject's performance at a synchronization value by averaging their performance at all frequencies tested for that synchronization value. Meanwhile, the SNR term was represented by the expected (average) value of each CCP value normalized by $\sigma_P$ (the variance of a subject's performance scores across all frequencies at a synchronization value) over all subjects.

$$CCS = ste(CCP) * \log_2\left(1 + E\left(\frac{CCP}{\sigma_P}\right)\right) \qquad (10)$$

The hypothesis was that an increase in the synchronization gain would not only improve CCP, by assisting the subject, but also improve the assessment's ability to

differentiate between subjects, via an increase in CCS. It would do this by reducing subject variance relative to the test bandwidth.

# 4 Results & Observations

The first two specific aims of this thesis included the development of an instrumented balance platform and the design of an assessment protocol that challenged the subject's ability to regulate self-perturbations. Because these two aims were addressed while explaining the methods of this work, the results outlined in this Section focus on achieving the 3rd specific aim specified in Section 1.5. This aim sought to evaluate the promise of adaptive balance assessment by analyzing the effect of synchronization. The first step towards this aim was therefore to investigate whether synchronization effectively assisted subjects during the assessment. The second step towards achieving this aim was to evaluate the ability of synchronization to improve CCS. The results in this section show that neither of these two relations could be confirmed within the scope of this work. This section consequently details a set of preliminary diagnostics which may help in understanding this outcome and periodically makes observations of their implications.

## 4.1 Preliminary diagnostics

The raw data confirmed several observations made by subjects during testing. As shown in the top row of Figure 4.1, there were times where subjects were able to track the instructions consistently throughout a trial. Other times, as seen in the bottom row of Figure 4.1, subjects had difficulties matching the instructions and employed various strategies to compensate. These strategies included altering their frequency of oscillation, undershooting an oscillation, or stopping until the instructions returned to their angular position. Appendix B provides more sub-trials, like the ones in Figure 4.1, where tracking went relatively well and when these compensatory strategies were used.

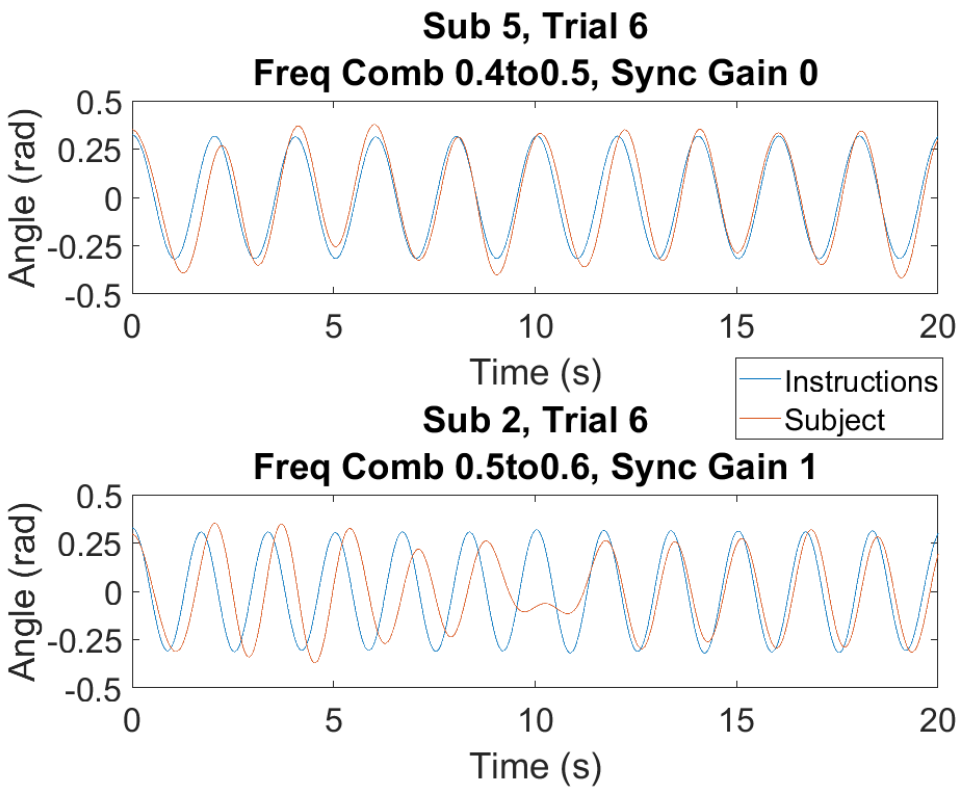



**Figure 4.1. Example angular position data. The top plot is an example of a subject successfully tracking instructions, while the middle section of the bottom example shows the subject employing a stopping technique to catch up with the instructions.**

### 4.1.1 Threshold-based results

Within the threshold-based measures, only a few trends were observed. The mean error and median error plots, as shown in Figure 4.2, consistently had a higher error after the threshold was met for higher frequencies. This indicates that after a threshold was met, subjects produced more error at higher frequencies. The delay error consistently exhibited larger delays before going below the threshold at higher frequencies, but, as shown in Figure 4.3, the y-axis revealed that the thresholds triggered relatively early after a frequency change. Further inspection revealed that these thresholds always triggered

within the first period. As shown in Appendix A, the remaining threshold measures produced no trends.

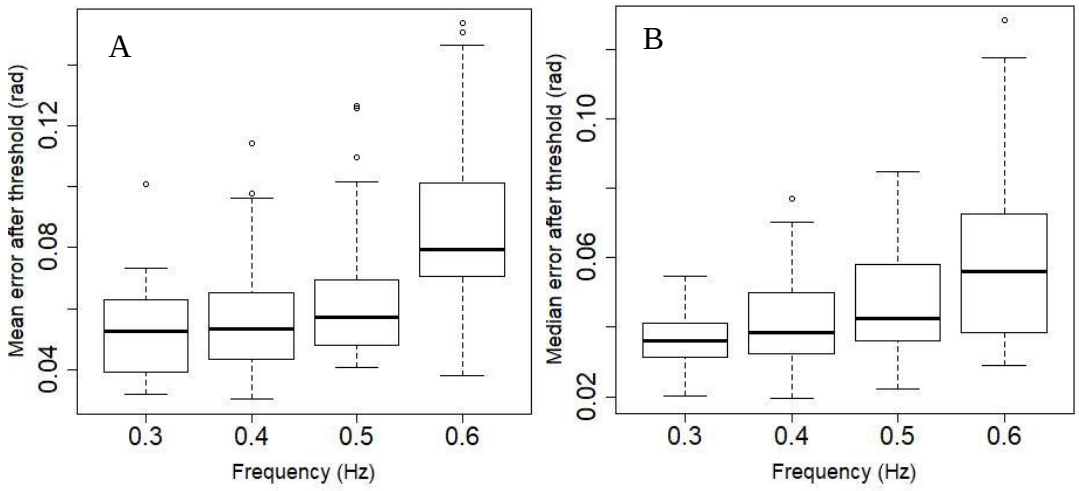



**Figure 4.2. A) Effect of frequency on the mean error threshold time for a synchronization value of 0. B) Frequency's effect on the median error threshold time for a synchronization value of 2.**

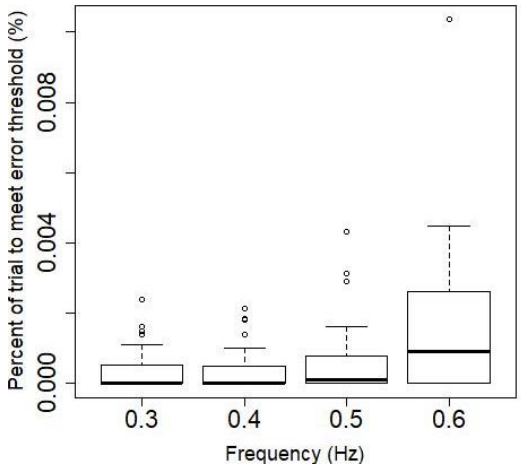



**Figure 4.3. Effect of frequency on percent of the sub-trial it took to get below the threshold for delay error for a synchronization of 4.**

### 4.1.2 Compliance-based results

Within the four fit-based SNR error measures, two measures evaluated the instructions' angular position, which indicates how much the instructions had to change

due to synchronization, and the other two evaluated the subject's angular position, which partially indicated a measure of subject performance (not accounting for bandwidth as discussed in Section 3.5.3). The mean SNR of the instructions, as shown in Figure 4.4A for one of the synchronization values, indicated a linear trend in which lower SNR values occur at higher frequencies. This trend was consistent across all synchronization values. The instruction area SNR displayed two trends. Holding synchronization constant, as shown in Figure 4.4B as an example, consistently had a parabolic trend suggesting that a target frequency of 0.5 Hz produced the highest SNR results for the instructions. Meanwhile the plots of constant frequency, as shown in Figure 4.4C, consistently had an exponential trend in which higher synchronization values produce exponentially lower SNR values. The other two SNR measures analyzed the subject data. The boxplots for mean SNR at a constant synchronization value, as shown in Figure 4.4D, consistently had a linear trend indicating that higher frequencies produced lower SNR values. Meanwhile, the boxplots for area SNR at a constant synchronization value, as shown in Figure 4.4E, consistently had a parabolic trend indicating that SNR values peaked closer to 0.5 Hz.

Finally, though most boxplots for $R^2$ did not produce any noticeable trends, those with constant synchronization values, as exemplified in Figure 4.4F, indicated that higher frequencies produced lower mean and higher variability results. This trend was consistent across all levels of synchronization.

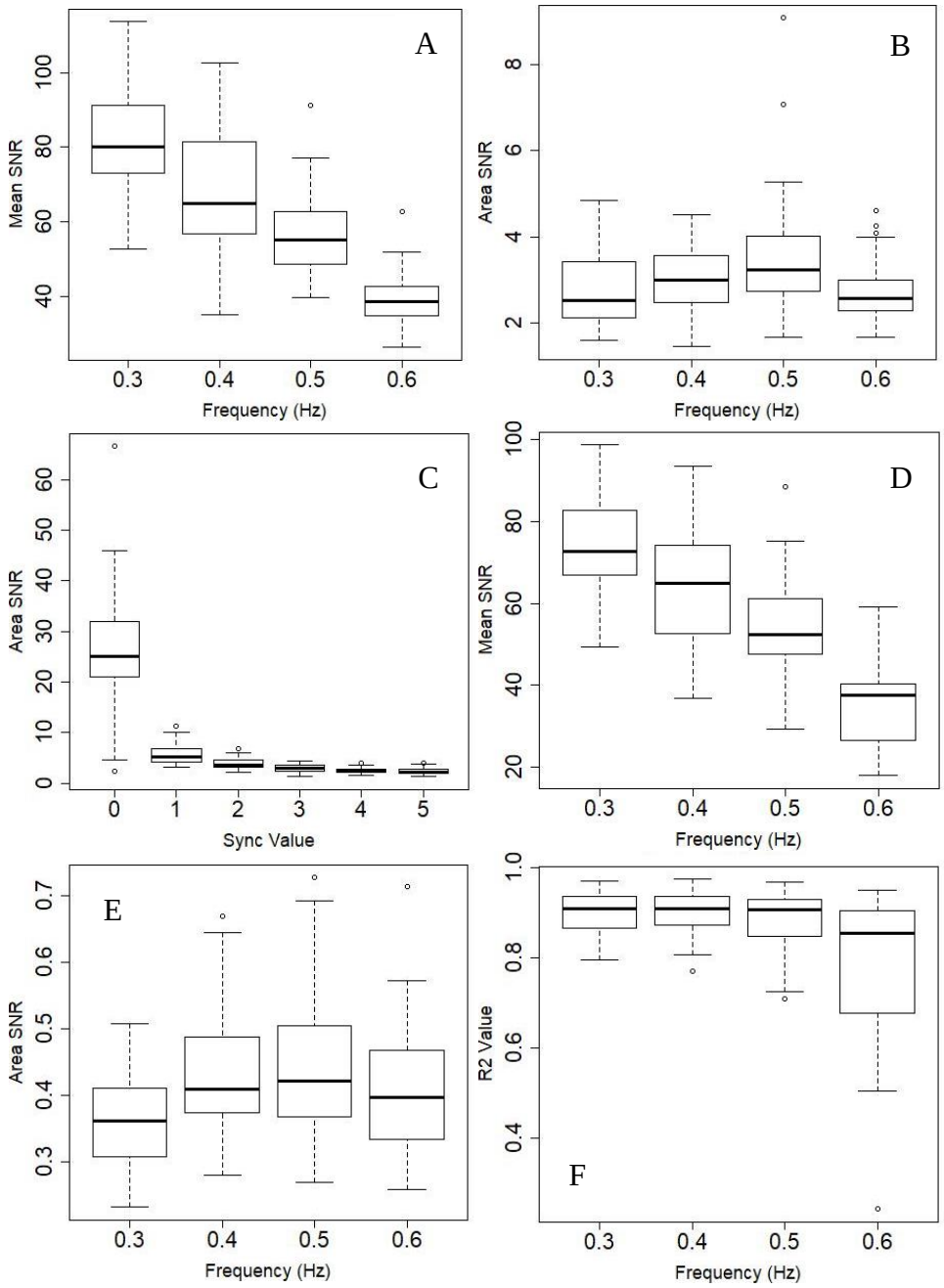



**Figure 4.4. A) Frequency's effect on instruction mean SNR for a synchronization value of 2. B) Frequency's effect on instruction area SNR for sub-trials with a synchronization value of 3. C) Synchronization's effect on instruction area SNR for sub-trials at a target frequency of 0.4 Hz. D) The subject's mean SNR for sub-trials at a synchronization value of 2. E) The subject's area SNR for sub-trials at a synchronization value of 4. F) Frequency's effect on coefficient of determination for a synchronization value of 2.**

### 4.1.3 Observations

Although several trends appeared, many of these measures had flaws worth noting. One of the defining characteristics of synchronization is an exponential convergence of all systems. But, as shown in an example in Figure 4.5, error measures did not exclusively decrease over time. It was expected that there would be some oscillations in the error, but even the amplitude of these oscillations did not strictly decrease over time. Therefore, the two oscillators, the instructions and the subject on the balance platform, did not exponentially converge as expected in a synchronizing system. Because of this unexpected outcome, the threshold-based measures did not function as predicted.

Thresholds were chosen by evenly spacing four values between the average maximum amplitude and the average minimum amplitude. Since the amplitude instructed to subjects was constant, this averaging was taken across the complete set of data. Data presented represent the medium threshold as results were similar for all four values. The threshold-based measures that required the error to drop (but not stay) below the threshold, were frequently triggered before the first quarter of the first period (as shown with delay in Figure 4.3). Meanwhile, the threshold measure that required the error measure to stay below the threshold frequently remained untriggered until the last period. Exploration of these results revealed that the selection of different threshold values did not fix these issues, either demonstrating similar behaviors, triggering too easily, or not at all.

Despite the lack of an expected trend, these results can still offer some insight on subject behavior. As there was little tolerance or ability to match the higher frequency instructions, these prompts produced more error, as shown Figure 4.4D. On the other hand, a looser tolerance appears to highlight that 0.5 Hz was closer to the ideal frequency as it

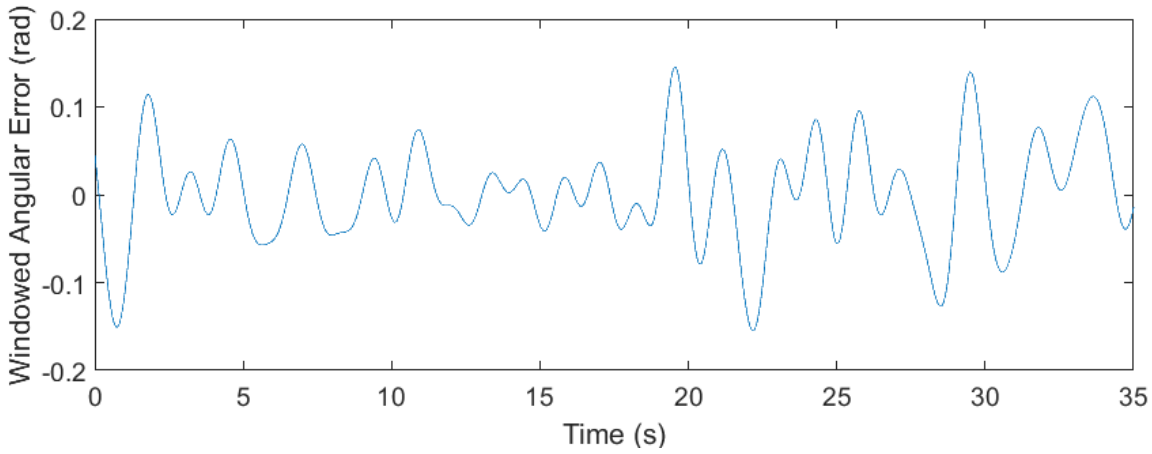



**Figure 4.5. Windowed angular error over time demonstrating the threshold issue: the amplitude does not decay over time.**

produced the highest SNR as shown in Figure 4.4E. Meanwhile, the instructions' SNR data agreed with these two previous findings and pointed out that when synchronization was applied, it did not remain at the target frequency (as shown in Figure 4.4C). This last observation corroborates that the system did indeed deviate from the original target frequency as it attempted to synchronize.

## 4.2 Channel capacity for subject performance (CCP)

### 4.2.1 Results

When examining the boxplots of the results of CCP, a positive trend between frequency and CCP was found, as shown on the left side of Figure 4.6. Other relationships, like that between CCP and synchronization, as shown on the right side of Figure 4.6, displayed no real trends.

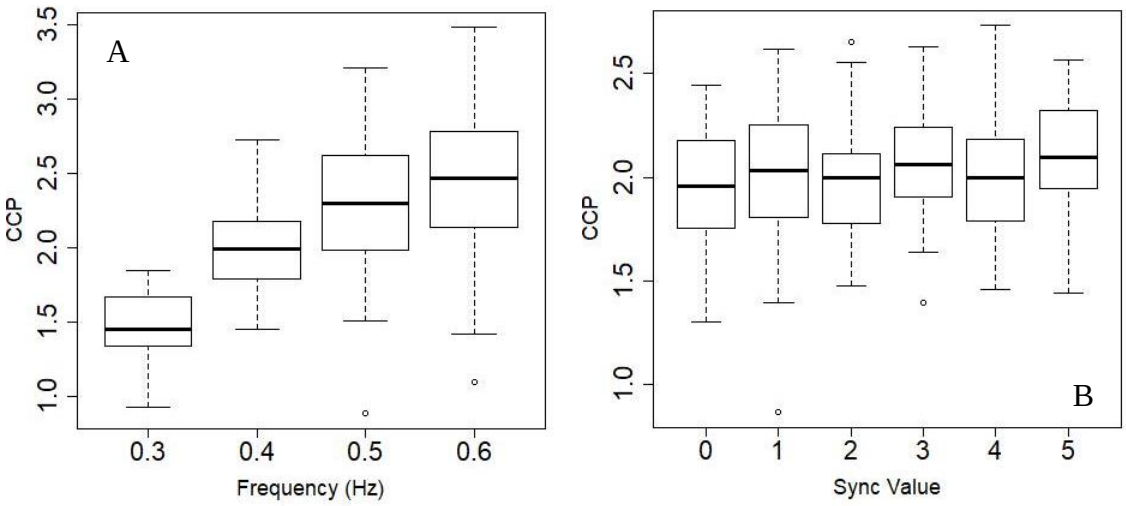



**Figure 4.6. A) Frequency's effect on CCP for a synchronization value of 4. B) Synchronization's effect on CCP for a target frequency of 0.4 Hz.**

### 4.2.2 Observations

To apply adaptive synchronization as discussed in Section 2.2.3, it must be demonstrated that CCP and synchronization increase and decrease together. The boxplots which examined this relation (see Figure 4.6B as an example), however, demonstrated overlapping distributions with no clear trends. Because no trend was present in this research, conclusions may not be drawn about whether synchronization influenced performance. Meanwhile. plots of CCP against frequency did demonstrate the direct effect of bandwidth on channel capacity as shown in Equation 9.

## 4.3 Channel capacity for the effect of synchronization (CCS)

### 4.3.1 Results

Contrary to expectations, Figure 4.7 shows that CCS did not increase with synchronization. No meaningful statistical analysis could be performed on the CCS values

presented in Figure 4.7 because each point was directly calculated using all subjects. Consequently, a subdivision method was explored from which an idea of the variance could be computed.

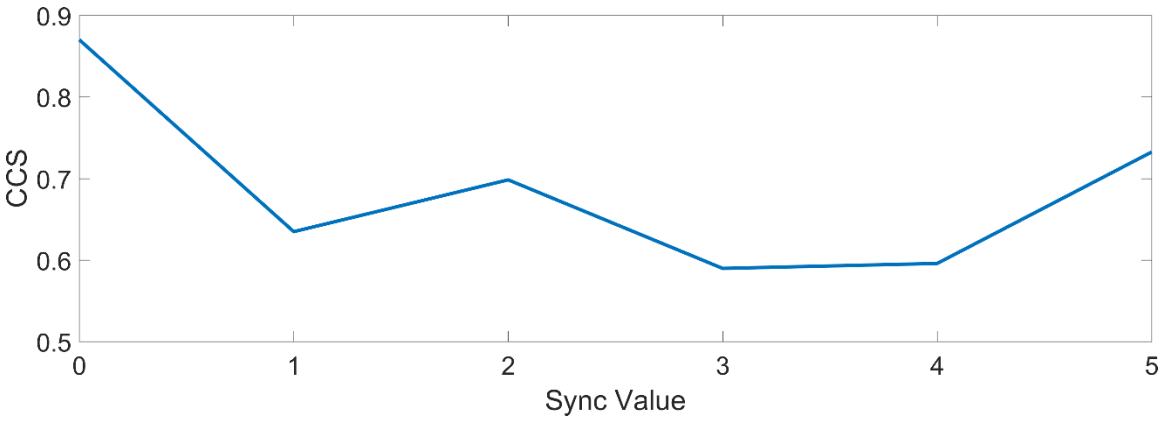



**Figure 4.7. Synchronization's effect on assessment precision (CCS).**

To obtain an idea for the spread of the data, the 19 subjects were divided randomly into four groups (three groups of 5, and one of 4). This produced 4 data points for each synchronization value, enabling the results shown in Figure 4.8 to be computed. Given these inconclusive results, no additional statistical analyses were performed.

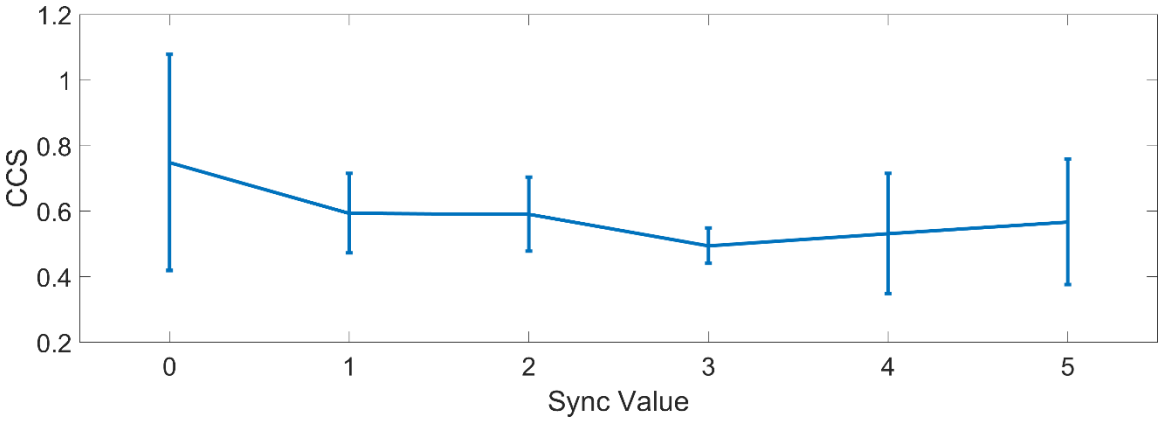



**Figure 4.8. Synchronization's effect on assessment precision (CCS). Data were subdivided into 4 subgroups to create the error bars.**

### 4.3.2 Observations

As explained in Section 3.5.4, CCS quantifies the repeatability of subject performance measure CCP. Therefore, Figure 4.7 and Figure 4.8 suggest that CCP was repeatable for midrange synchronization values, but was otherwise relatively inconsistent. As there is no assistance at a synchronization value of 0, it was expected that this value would receive the highest inconsistency, as observed. The lack of a discernable slope, however, suggests that CCS was largely independent of synchronization. As a result, further work is required before it can be conclusively recommended that future studies apply adaptive synchronization for the proposed purpose.

# 5 Discussion

The results presented in the previous section brought forth three important conclusions. The preliminary diagnostics suggested that the instructions and subject were not synchronizing as they were not exponentially converging over time. Consequently, the CCP analysis could not demonstrate that increasing synchronization gain improved assistance to the subject during testing. Finally, the CCS analysis could not demonstrate that adaptive synchronization would produce a higher precision in subject outcome/performance measure CCP. The following Sections discuss the theory that should have supported this effect, reviews the limitations of the experiment as it was designed, and from this discussion provides suggestions for future work.

## 5.1 The role of feedback

### 5.1.1 Within-trial feedback

As discussed in Section 2.2.1, part of within-trial feedback provided the subject with information on their position to compare it with the instructions, using a monitor, throughout the experiment. This feedback was applied to make the experiment more akin to the dynamic and self-perturbed falls that are the most common falls in older adults as the subjects would have similar information available in those situations.

However, this also brought with it some benefits against disturbances with a duration smaller than a single trial. These disturbances included: short-term subject distractions (such as construction noises and knocking on the experiment room door), small periods of computer lag, short-term subject fatigue, and more. Though it was uncertain

how much of an effect these examples of disturbances had, or could have had with adaptive synchronization, each one was present to some degree in this experiment.

### 5.1.2 Adaptive synchronization

As discussed in Section 3.3, the long-term plan of this research would see feedback applied to the performance/outcome measure (CCP) in the completed balance assessment. This feedback would adjust the synchronization gain adaptively, either continuously or between trials, altering the user assistance appropriately to the user's abilities. When the desired CCP matched the user's CCP, the assistance required to do so would be the output used by the clinician to evaluate the subject.

The current experiment could not apply this feedback method however, as it would have assumed that the synchronization gain and CCP would always increase and decrease together. Instead of applying feedback adaptively between trials, the experiment collected data on the effects of synchronization by observing CCP at several different synchronization gains. The results however, as discussed in Section 4.2.2, could not confirm the fundamental benefit of synchronization to warrant the use of adaptive synchronization.

If the hypothesized effect of synchronization was later confirmed and applied to the balance assessment using adaptive feedback, it would boast two main advantages: (1) it would linearize the system and (2) would make control more robust to perturbations. However, it was also important to note that using this feedback also comes with three main disadvantages: (1) a risk of making the system unstable, (2) a risk of emphasizing noise, more than the signal of interest, and (3) a risk of having noise further corrupt the output

measures. Though the experiment conducted in this thesis provided inconclusive results, it was still worthwhile to evaluate these advantages and disadvantages to theoretically assess the promise of this balance assessment for future studies.

### 5.1.3 Advantages

The first expected advantage of using feedback adaptively between trials was linearization. As mentioned in Section 2.2.3, the system transfer function (Equation 7) was linearized when $|kP| \gg 1$ and, as a consequence, the numerator and denominator become roughly equal. This was ideal for adaptive synchronization as the final design intends to have CCP (the output of the transfer function) match the desired CCP (the input of the transfer function) and collect the assistance required (synchronization gains) as the measure used by clinicians to evaluate the subject's balance. Furthermore, since CCP was calculated with a nonlinear formula (Equation 9), its effect was always linearized as the system's plant was present on both the numerator and denominator.

System robustness to disturbances is also a product of feedback and constitutes the second advantage for adaptive synchronization in a completed balance assessment. After every trial, the following trial's synchronization gains would be altered to assist the subject in closing the gap between the performance measure and the desired performance measure. This gap may be created due to the subject's ability (or inability) to perform the test, any additional disturbances, or other sources of noise.

Disturbances that may have affected subjects for more than one trial in duration, and therefore could have benefited from adaptive synchronization, include auditory distractions (construction noise, interruptions, talking), subject fatigue, boredom,

vibrations from construction equipment, stiff mechanical joints, balance platform flexion, calibration errors (between the on-screen instructions and the balance platform), subject gradual improvement, and more. Note that subject fatigue was tempered with breaks after every 3rd trial (as explained in Section 3.4.2) to maintain a low level of error from this source.

### 5.1.4 Disadvantages

Aside from theoretical complexity and monetary cost, the main disadvantages of feedback in closed-loop systems include a risk of making the system unstable, a risk of emphasizing noise more than the signal of interest, and a risk of having noise further corrupt the output measures. To assess all these disadvantages, it will help to first calculate the system's "gang of four" which comprises sensitivity, complimentary sensitivity, load sensitivity, and noise sensitivity.

These forms of sensitivity were derived by relating outputs of interest ($x$, $y$, and $u$) to inputs of interest ($r$, $d$, and $n$). Assuming a simplified control diagram for this balance assessment as represented in Figure 5.1, the gang of four may be calculated using Equations 11-14.

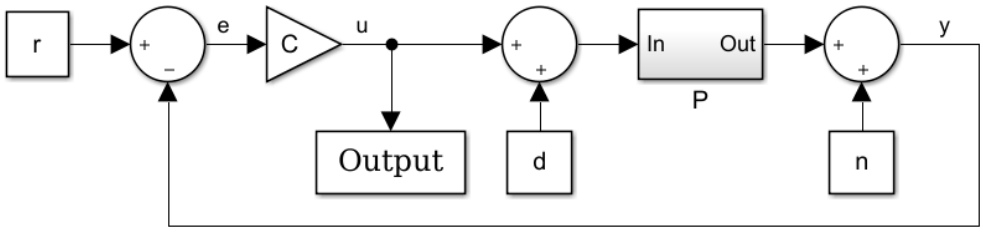



**Figure 5.1. A conceptual control diagram for the finalized balance assessment with adaptive synchronization. $C$ represents the controller which produces a synchronization gain based on the desired CCP $r$ and the actual CCP $y$, $d$ is the disturbances, $P$ is the user's ability, and $n$ is the noise.**

$$Sensitivity = \frac{1}{1+CP} \quad (11)$$

$$Complimentary\ Sensitivity = \frac{y(s)}{r(s)} = \frac{CP}{1+CP} \quad (12)$$

$$Load\ Sensitivity = \frac{y(s)}{d(s)} = \frac{P}{1+CP} \quad (13)$$

$$Noise\ Sensitivity = \frac{u(s)}{n(s)} = \frac{-C}{1+CP} \quad (14)$$

In this control diagram, $r$ represents the desired outcome measure, $C$ is the controller, and $P$ is the subject's ability. Other variables denote different steps within the control diagram and helped create the gang of four. These variables include $e$ for error, $u$ for input, $d$ for disturbance, $n$ for noise, and $y$ for the performance measure (CCP).

Equation 11 demonstrates what was added to the system transfer function when adding a feedback loop. Equation 12 was the system's output over input transfer function and represents the most commonly assessed transfer function for stability. Meanwhile, instability may arise under the presence of disturbances, perturbations directly affecting the plant, and noise, errors present in measurements made on the plant output or trial abnormalities that can bias the data analysis. To assess the stability of a system in the face of disturbances, Equation 13 may be used. Likewise, to assess the stability of a system in the face of noise, Equation 14 may be used.

#### 5.1.4.1 Instability

The first disadvantage to consider was feedback making the system unstable. Employing linear controls analysis, there were many ways to investigate system stability using the complimentary sensitivity Equation (Equation 12). For example, one method was to use a pole-zero map to investigate the placement of poles (the values of s when factors

in the denominator turn to zero). If any of these poles occurred at positive real values on the s plane, the system would be unstable. So long as adaptive synchronization did not create a positive feedback loop, due to the effect of synchronization on CCP, this shouldn't be an issue. However, as noted in Section 4.2.2, the effect of synchronization on CCP is unclear. Therefore, no conclusions could be drawn on this relation.

#### 5.1.4.2 Sensitivity

The second possible disadvantage is a decrease in the assessment's sensitivity to the output with respect to the sensitivity of disturbances. Typically, balance assessments have been conducted in an open-loop setup as seen in Figure 5.2. This is identical to the closed-loop diagram with its variables in Figure 5.1, but there's no feedback loop.

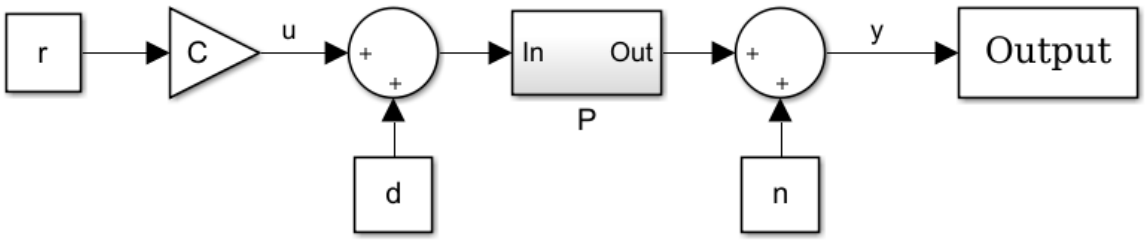



**Figure 5.2. Open-loop balance assessment. $C$ represents the controller which interprets the test input from $r$, $d$ is the disturbances, $P$ is the user's ability, $n$ is the noise, and $y$ is the output.**

For the sake of measuring sensitivity, if the variable $m$ represents the test's output, then for an open-loop system it would be equal to $y$ and for a closed-loop system it would be equal to $u$. Ideally, a balance assessment is sensitive to changes in the innate abilities of the subject and robust to disturbances. In other words, an assessment should aim for a high sensitivity to changes in the output with respect to changes in the subject's abilities ($\frac{\partial m}{\partial P}$) and a low sensitivity to changes in the output with respect to changes in the disturbances

($\frac{\partial m}{\partial d}$). To calculate and study these sensitivities, Equations 12-15 calculate these relations where the subscript "$OL$" is for the open-loop system and "$CL$" is for the closed-loop system. Figure 5.3 and Figure 5.4 show how disturbance and subject ability affect these sensitivities as systems apply a feedback loop while maintaining otherwise identical parameters.

$$m_{OL} = n + dP + rCP$$

$$\frac{\partial m_{OL}}{\partial P} = d + rC \tag{12}$$

$$\frac{\partial m_{OL}}{\partial d} = P \tag{13}$$

$$m_{CL} = \frac{rC - nC - dPC}{1 + CP}$$

$$\frac{\partial m_{CL}}{\partial P} = \frac{C(Cn - Cr - d)}{(1 + CP)^2} \tag{14}$$

$$\frac{\partial m_{CL}}{\partial d} = \frac{-CP}{1 + CP} \tag{15}$$

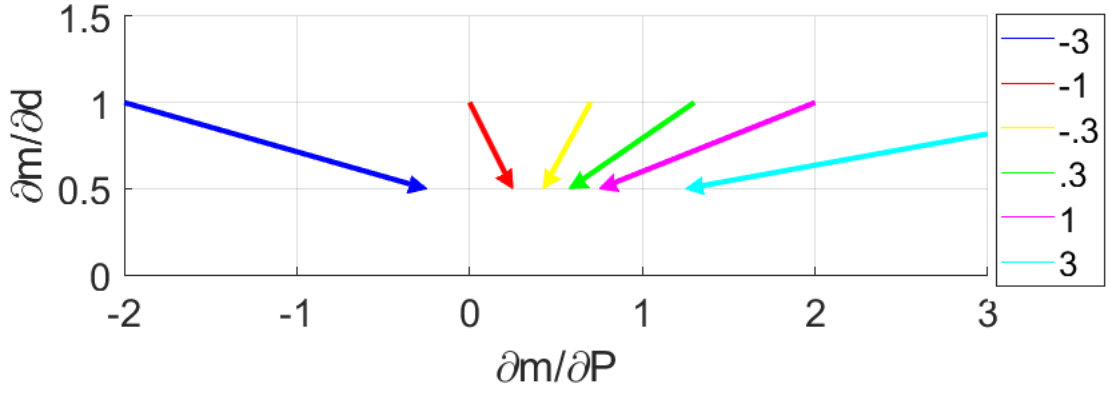



**Figure 5.3. Output sensitivity to disturbances. The legend lists six different disturbance values used. The arrows point from an open-loop to the closed-loop system equivalent while maintaining identical parameters.**

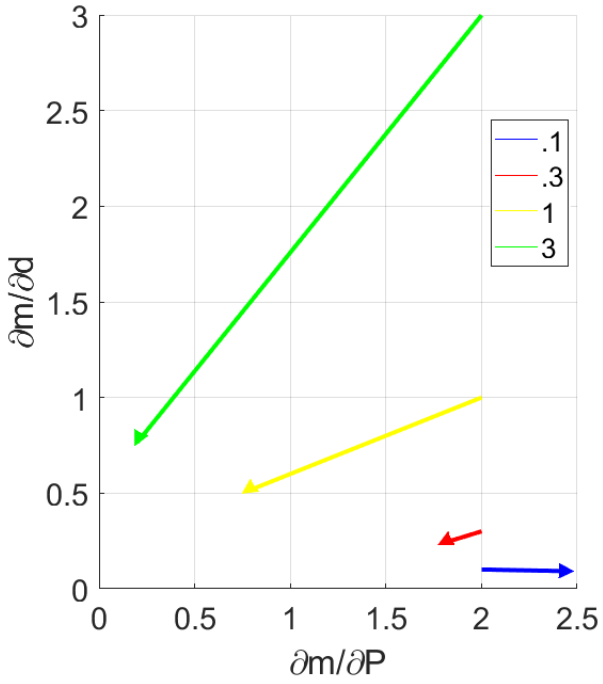



**Figure 5.4. Output sensitivity to subject performance. The legend lists four different performance values used. The arrows point from an open-loop to the closed-loop system equivalent while maintaining identical parameters.**

With respect to output sensitivity to disturbances, the controller gain has no effect in open-loop systems (Equation 13). For closed-loop systems (Equation 15), a high controller gain brings the sensitivity to unity. If $P$ is too small however, this sensitivity is amplified relative to this sensitivity in an open-loop system. With smaller controller gains, this sensitivity tends to 0.

With respect to output sensitivity to subject ability, a high controller gain simplifies the open-loop system Equation 12 to $rC$ and the closed loop system Equation 14 to $\frac{n-r}{P^2}$. Relative to the open-loop system, a closed-loop system's output sensitivity to subject ability is attenuated by a factor of $P^2$. With smaller controller gain values, the results aren't so clear.

It accordingly seems that using a closed-loop system with a large controller gain is not advantageous. The case becomes less clear for small gains and depends on the particular range of values for $r$, $d$, and $P$. In some cases the inclusion of a feedback loop has a stronger net-gain in terms of rejecting disturbances than the associated net-loss in terms of reducing sensitivity to subject ability. Of course, if the control signal is set too small, the sensitivity drops to 0. A balance must therefore be struck to make the most use of closed-loop feedback.

#### 5.1.4.3 Noise sensitivity

In any balance assessment, noise can be introduced during the trials and during the data analysis. By using the right data analysis techniques, noise may be filtered, to some extent, before producing the outcome measure. But if feedback is added to the system, the outcome measure with its noise would be reintroduced to the system. This third disadvantage to feedback could create further erroneous variations in the outcome measure, which affects the adjustments made to the synchronization gain, and may lead to instability. In certain cases, a system could go unstable due to the noise added to the feedback.

For the finalized balance assessment with adaptive synchronization, examples of noise likely included: unanticipated line noise, physical vibrations (to the mechanical joints, hall-effect sensors, or wires from a construction site outside), calibration errors, joint stiction (static friction), trial abnormalities that bias analysis, and more. It was unlikely that any of these noises could have destabilized the system however due to the subject. If a subject was theoretically able to follow the instructions relentlessly, then the oscillatory noise (like unanticipated line noise and physical vibrations) could have destabilized the

system. Realistically however, it was noticed that subjects would adopt other mechanisms if they couldn't follow the instructions. For example, when subjects couldn't keep up with instructions, they waited in place until the instructions came back to their current position and continued following instructions from there. Meanwhile, calibration errors due to drifting sensors would manifest as stuttering in the instructions, a reduced range of motion, and a constant angle bias. In this case, and in the case of joint stiction, subjects acknowledged the noise and attempted to maintain a smooth motion. Therefore, it was unlikely any of these could have caused an instability with adaptive synchronization.

## 5.2 Contributions

Although the observed results did not support the expected outcomes, this work represents a collection of tangible contributions towards the overall project goal of creating a balance assessment for older adults. These contributions include:

- The development of an instrumented balance platform designed to assess subject balance under dynamic self-perturbations. This development was guided by the need for a quick, affordable, objective, repeatable, and portable balance assessment. Section 3.1 presented the novel balance platform which was built and instrumented based on these guidelines.
- The foundations for a balance assessment test that challenged the subject's ability to regulate dynamic self-perturbations. Sections 3.2 to 3.4 outlined how data were processed, the model used for synchronization, and the test protocol designed and used to collect data from 19 young healthy adult subjects. Future work using adaptive synchronization in assessing the ability of older adults to handle dynamic self-

perturbations may build upon this testing methodology to create the final balance assessment.

- The evaluation of the potential of an adaptive balance assessment using synchronization which demonstrated that more work is required. The resulting data from the 19 subjects were analyzed using a framework of novel performance measures which integrated the effects of feedback and the tradeoff between bandwidth and SNR. While the results were inconclusive, as no significant trends could be observed, this work defined a starting point for assessing future balance studies with adaptive synchronization.

## 5.3 Limitations and Future work

Throughout the experiment there were several noted limitations and sources of error that may have played a role in the observed results. These limitations fall in one of several categories including: balance platform design, experiment design, and unintended disturbances.

### 5.3.1 Balance platform sensors

As described in Section 3.2, four hall-effect sensors were used to interpolate the platform's angular position using their voltage readings as shown in Figure 3.3. Several issues with the Hall-effect sensors arose during testing which required frequent recalibration. Unfortunately, recalibrating could not be performed while testing a subject, as it would make data inconsistent within a subject, and the procedure to recalibrate took over 30 minutes. In future iterations, researchers may want to consider different or

additional methods of detecting platform angles, such as potentiometers and/or inertial measurement units.

One issue with the Hall-effect sensors occurred because of the chosen interpolation method. Due to the aforementioned strategy of dropping sensors when they got to close to their troughs, stuttering frequently occurred when a sensor was dropped, and the remaining sensors did not agree on the angular position of the balance platform. In total, five subjects reported frequent stuttering near the angles indicated by the vertical black lines at the troughs of Hall-effect sensors 2 and 4 as shown in Figure 3.3. Another five subjects reported some difficulties matching the extremities of the instructions' swaying motion. Both of these problems were likely the product of a degraded calibration and/or of having to drop other sensors from the interpolation function as they reached voltages above 2400 mV. Finally, six subjects reported the overall range of motion to be biased to one side at times. This flaw was most likely due to drift in the sensors that may have been caused by a physical displacement of the sensors relative to the magnets in the linkages.

### 5.3.2 Balance platform mechanics

The first mechanical source of error may be due to sticking friction (a.k.a. stiction) near one of the joints. Though only explicitly noted by the last subject, an inspection of the platform revealed that the wooden platform had shifted closer to the linkages, causing them to rub together. Figure 5.5 indicates one location where this rubbing occurred, highlighted by a red circle. Some improved mechanical techniques could be used to ensure less shifting in the platform, such as a washer and a sunken hole for the head of the bolt which fastened the wooden platform to the metal structure supporting it.

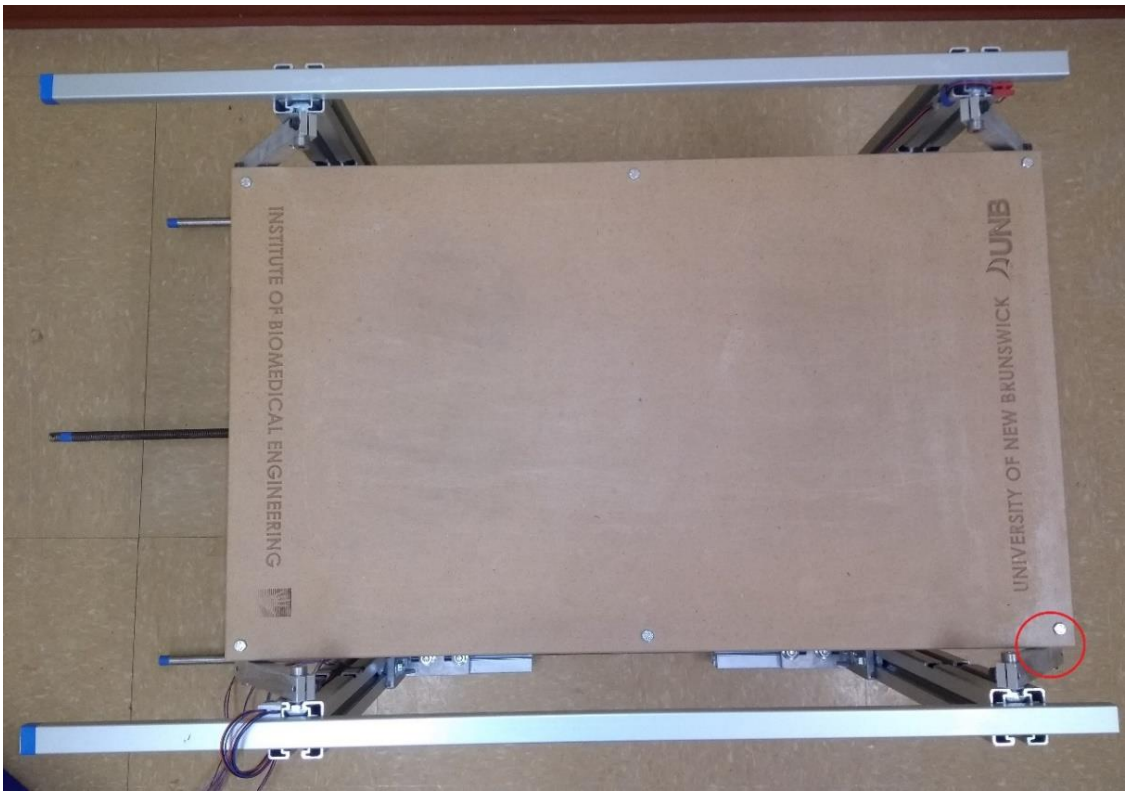



**Figure 5.5. Balance platform view from above. The red circle indicates a location where unintended friction occurred between the wooden platform and the linkage.**

Meanwhile, the one-dimensional design of the platform comes into question as the second balance platform limitation. The balance platform used in this study sways in only one direction to simplify the analysis of the data. However, this reduction in available data may be detrimental to the data analysis. Moreover, a few subjects mentioned they adopted methods to gain better control on the balance platform such as crouching or using their toes more. Such methods may not have worked on a platform that could sway on both axes. Future studies may want to adapt the platform to sway in both directions to better simulate real balance posture and to avoid reducing the attention needed for the balance assessment since older adults require more attention when maintaining balance than younger adults [26].

### 5.3.3 Experimental design

The chosen experimental design may have also introduced problems. Despite being instructed to place their feet at specific locations, a few subjects adapted their knee angle and the weight distribution in their feet to gain better control of the balance platform throughout the experiment. Because subjects only adapted these methods part way through the experiment, this may have affected some trials and not others. Future studies could specify knee angle and weight distribution to subjects to avoid variance presenting itself in this manner.

Meanwhile, the animation in the GUI, shown in Figure 3.4, was redundant in information. Several subjects noted that with the point of rotation being at the center of the black and green line, they could just concentrate on one side of the animation and receive all the information they needed to know to follow the instructions. It remains unclear if such a reduction in redundant information would help subjects perform better, but may be something future studies could investigate.

Given the simplified view that synchronization is two parallel applications of feedback, it may have been beneficial to determine the difference between a test with no feedback, one form of feedback, and two forms of feedback. For example, no feedback could have had a user train at different frequencies and then asked to later replicate them, one form of feedback could have had visual feedback on a monitor, but synchronization would be turned off, and two forms of feedback could have included synchronization. By adopting this experimental protocol, the effect of feedback in this scenario may have been clearer.

Meanwhile, despite efforts to stimulate subjects with mandatory breaks at every 3rd trial, half the subjects noted being fatigued (mentally and/or physically) by the last 2 of 12 trials. This was likely due to the repetitiveness of the experiment. Future studies may be able to better engage the subject by gamifying the experiment through practices such as providing a performance measure at the end of each trial or giving an imaginative purpose to the swaying motion. Future successful integration of online synchronization adaptation (rather than discrete fixed length and synchronization level trials) could also reduce the length of the test and improve engagement.

Finally, the method chosen for data analysis could be improved. If a subject did not match the instructions perfectly during testing, they would lose score as measured by CCP. This was typically a good method, but consider a scenario where a subject follows instructions, has a one-off mistake like in Figure 5.6, and then continues to perfectly follow instructions. Depending on how much the subject deviated from the instructions during this one-off mistake, the CCP score will decrease significantly. In scenarios such as these, it is possible that some form of outlier removal could avoid penalizing their CCP score, as it may not be reflective of their real balance ability.

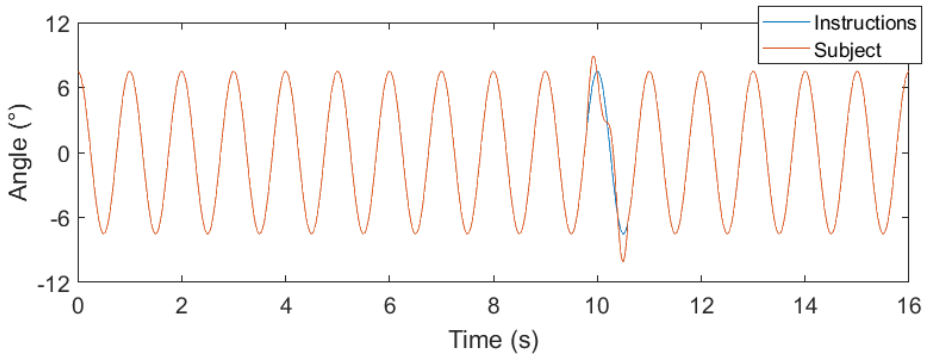



**Figure 5.6. A theoretical example of a subject making a one-off mistake in following the oscillatory motions of the instructions.**

### 5.3.4 Outside sources of error

With regards to unintended disturbances, it is possible that noise and vibrations due to ongoing building construction may have distracted some subjects. Though measurements of this noise do not exceed 75 dB (comparable to a toilet flush, a shower, a washing machine, or a vacuum cleaner), a few subjects did point out that the noise was distracting at times. However, due to location requirements (high ceiling with harness attachments) and availability, this noise was unavoidable within the time period of the study. Other outside sources of error may have included interruptions during testing from persons knocking on the lab door or subjects repeatedly speaking to the test administrator during testing (despite being told not to do so).

In future studies, should construction noise be an issue again, noise cancelling headphones could be used to eliminate some of the unintended noises. Meanwhile, signs should be placed on the door to avoid interruptions.

# 6 Conclusions

This thesis sought to take the first step in creating a balance assessment for older adults: an age group that falls are most frequently due to dynamic self-perturbations. To reduce the impact of movement noise in balance assessments, given an already limited range of motion in older adults, the promise of adaptive synchronization was investigated. This research was guided by the 3 specific aims outlined in Section 1.5:

- The first specific aim pursued the development of an instrumented balance platform. This development was guided by information in Section □, whereas Section 3.1 presented the novel balance platform that was built and instrumented based on this information.
- The second specific aim was to design an assessment protocol that challenged the subject's ability to regulate dynamic self-perturbations. Section 3.2 to 3.4 outlined the protocol designed and used to collect data from 19 young healthy adult subjects.
- The third specific aim sought to evaluate the promise of adaptive balance assessment by analyzing the effect of synchronization. The resulting data from the 19 subjects was analyzed using a framework of novel performance measures. While the results were inconclusive, as no significant trends could be observed, this work defined a framework for assessing future balance studies with adaptive synchronization.

Future studies posing similar experiments should consider several sources of error and limitations encountered in this experiment, including the process of comparing no feedback to one form of feedback and two-sided feedback (synchronization). Additionally, this three-stage investigation of synchronization could be applied to a much simpler test

with fewer mechanical parts in order to focus on synchronization's effect on human movement.

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# A. Appendix: threshold preliminary data (medium threshold)

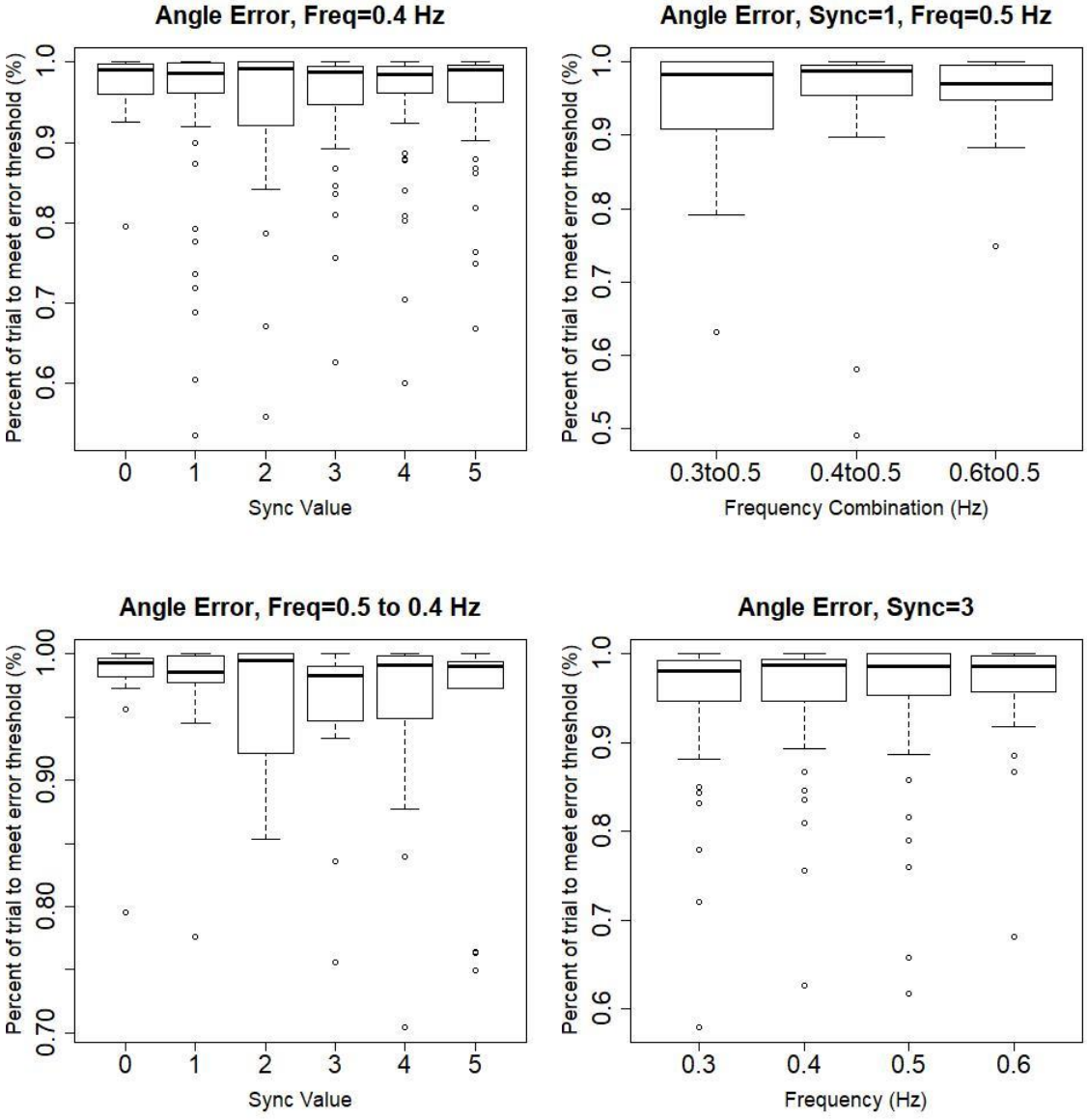



**Figure A.1 Angular error – staying below the threshold**

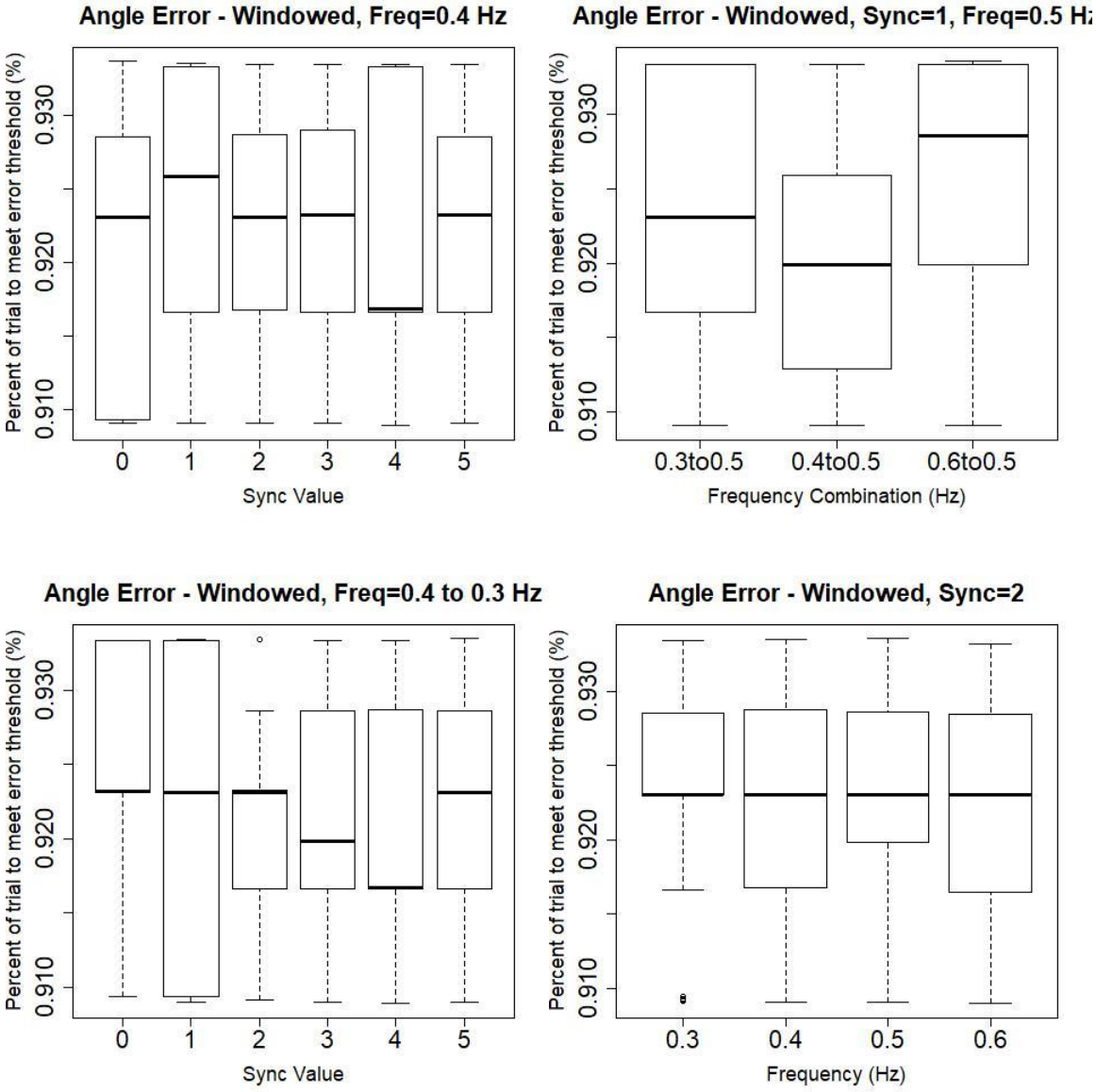



**Figure A.2. Windowed angular error – staying below the threshold**

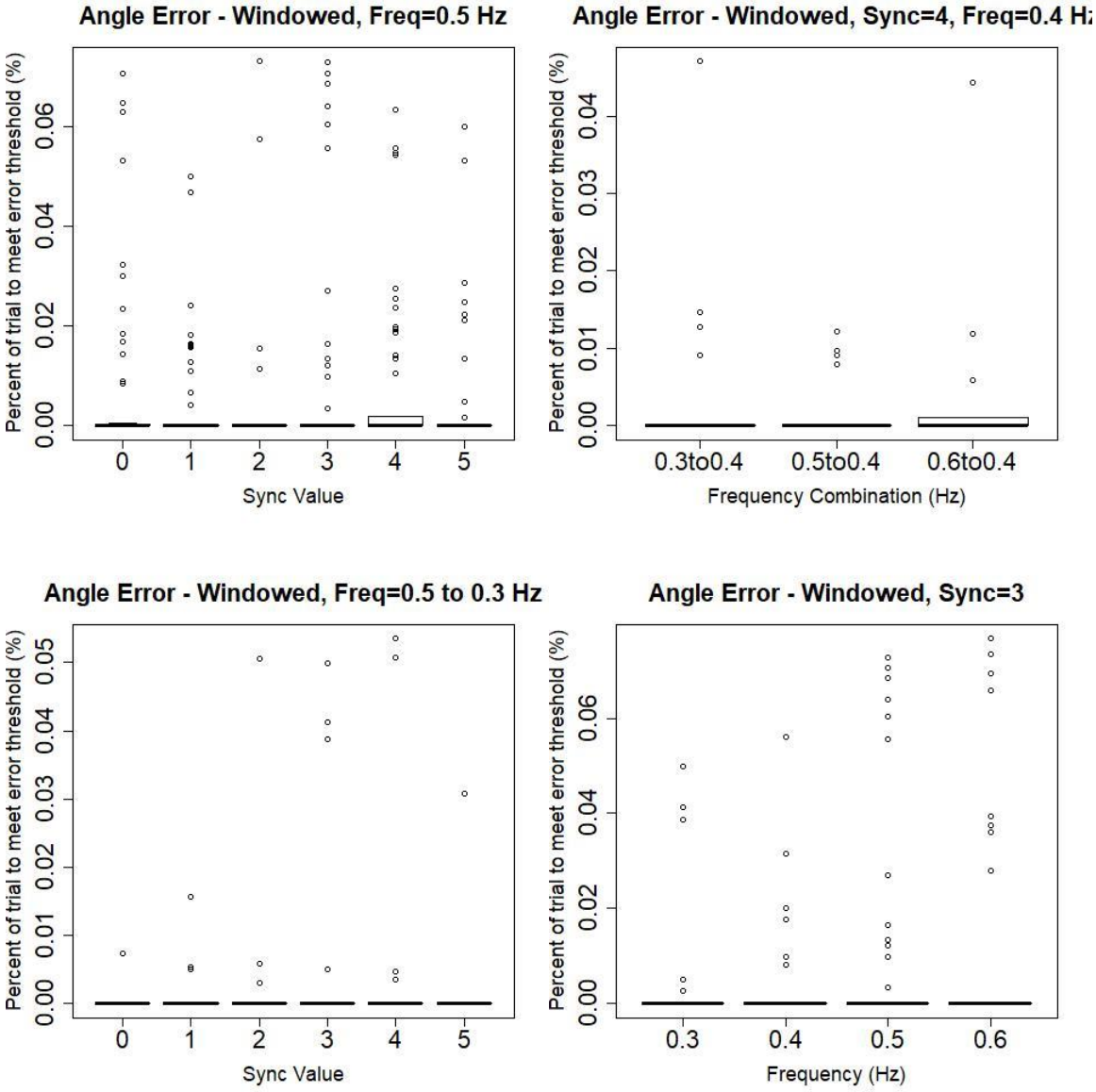



**Figure A.3. Windowed angular error – going below the threshold**

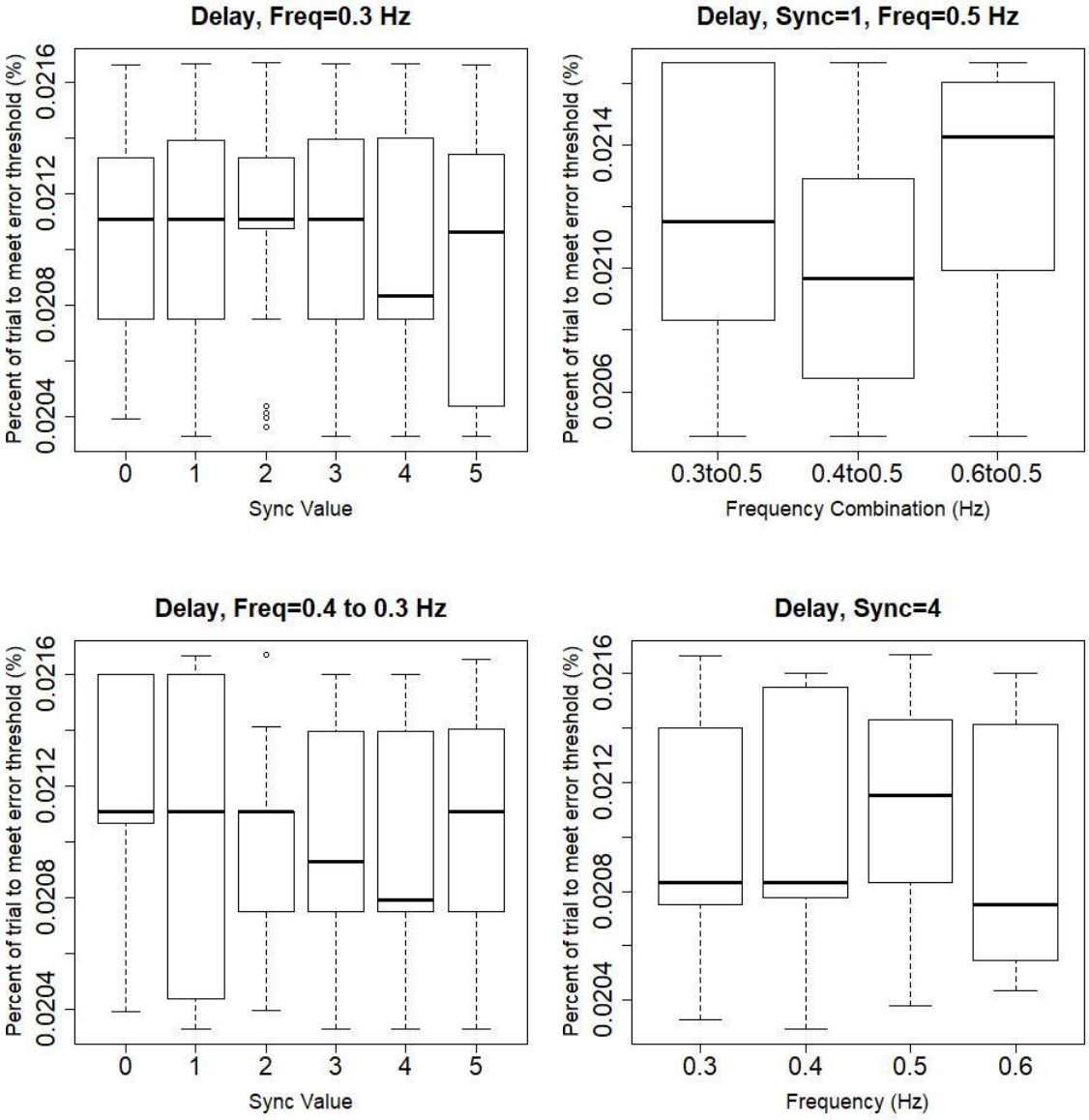



**Figure A.4. Delay – staying below the threshold**

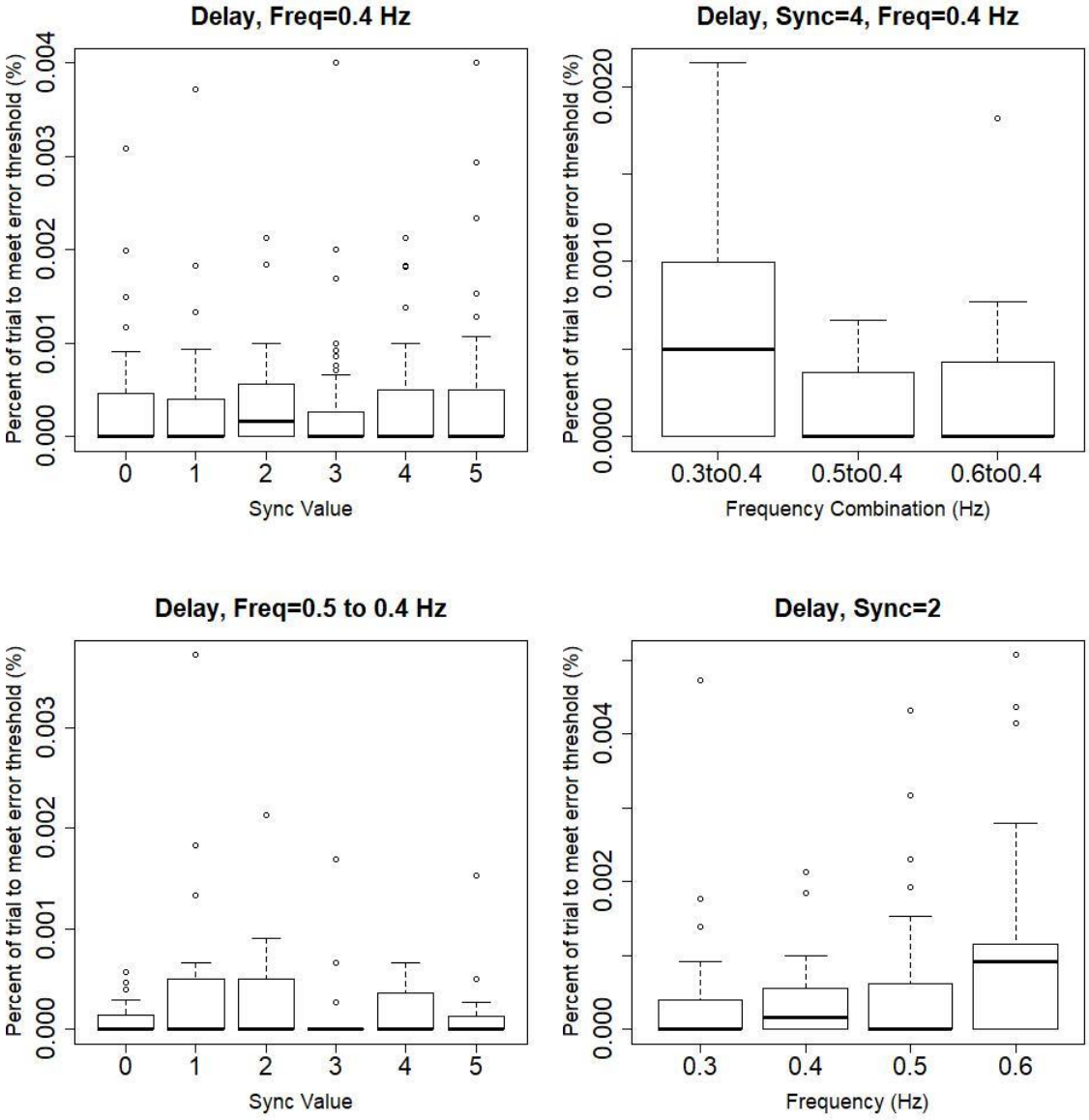



**Figure A.5. Delay – going below the threshold**

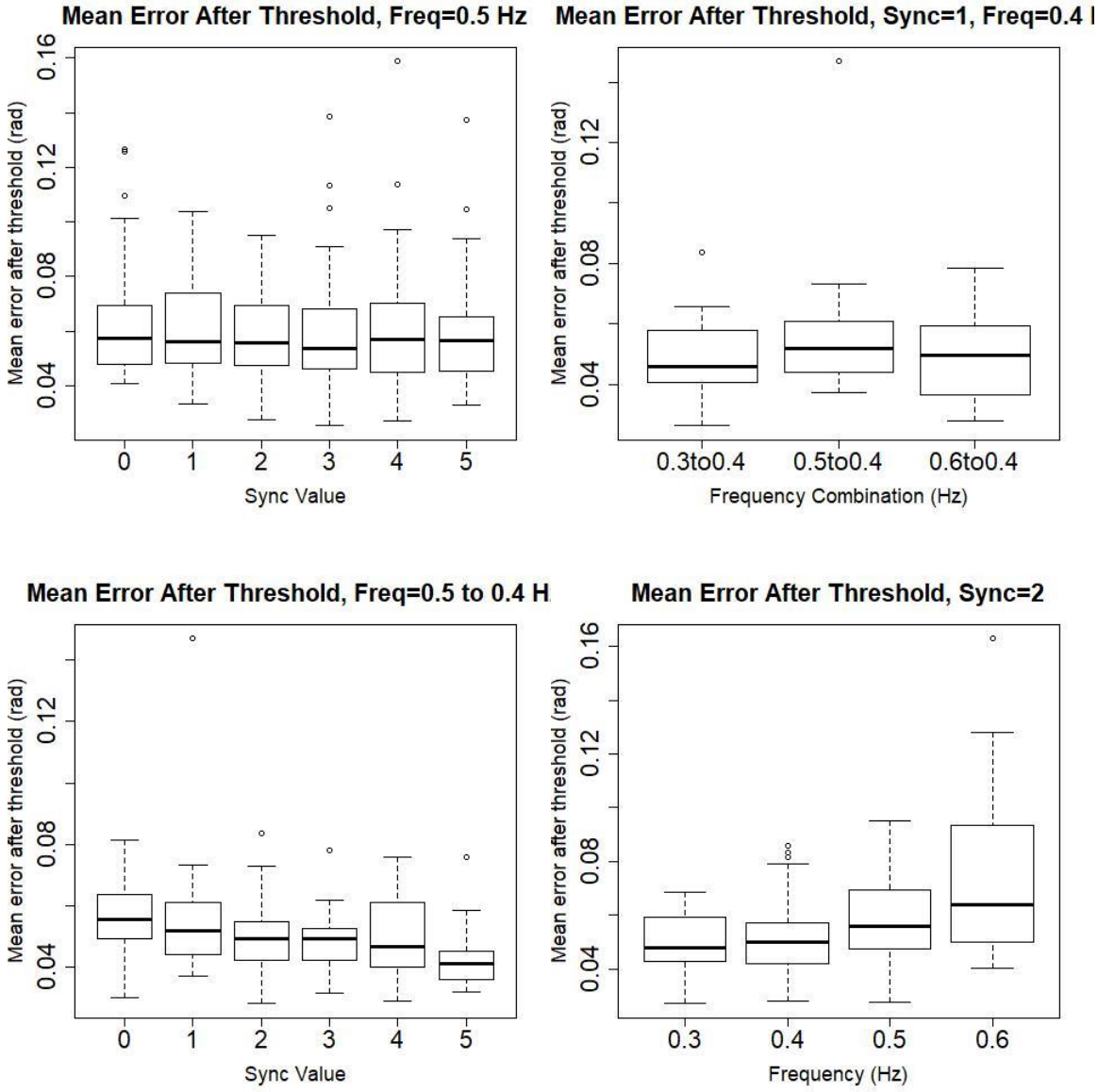



**Figure A.6. Mean error**

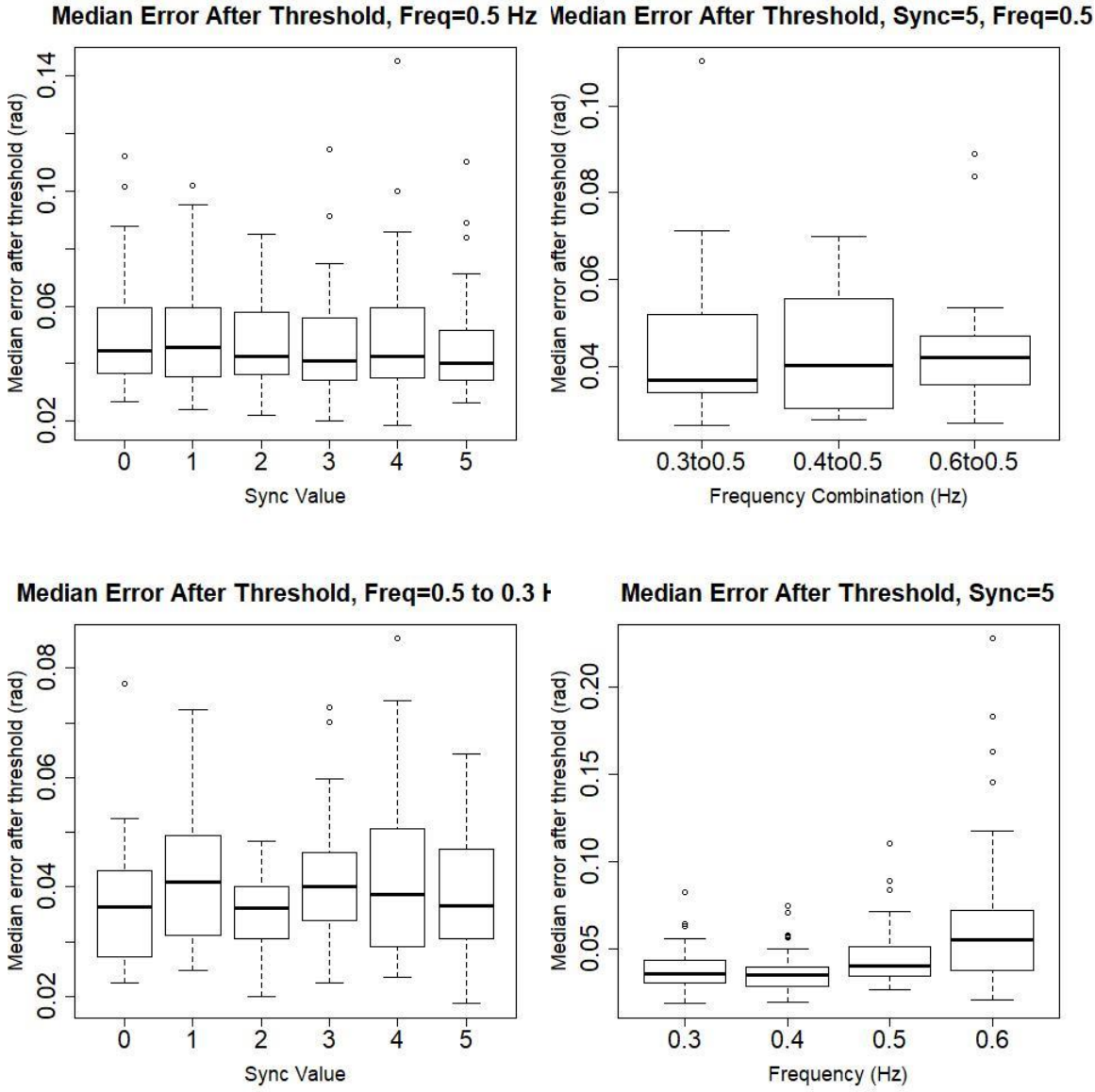



**Figure A.7. Median Error**

# B. Appendix: sample data

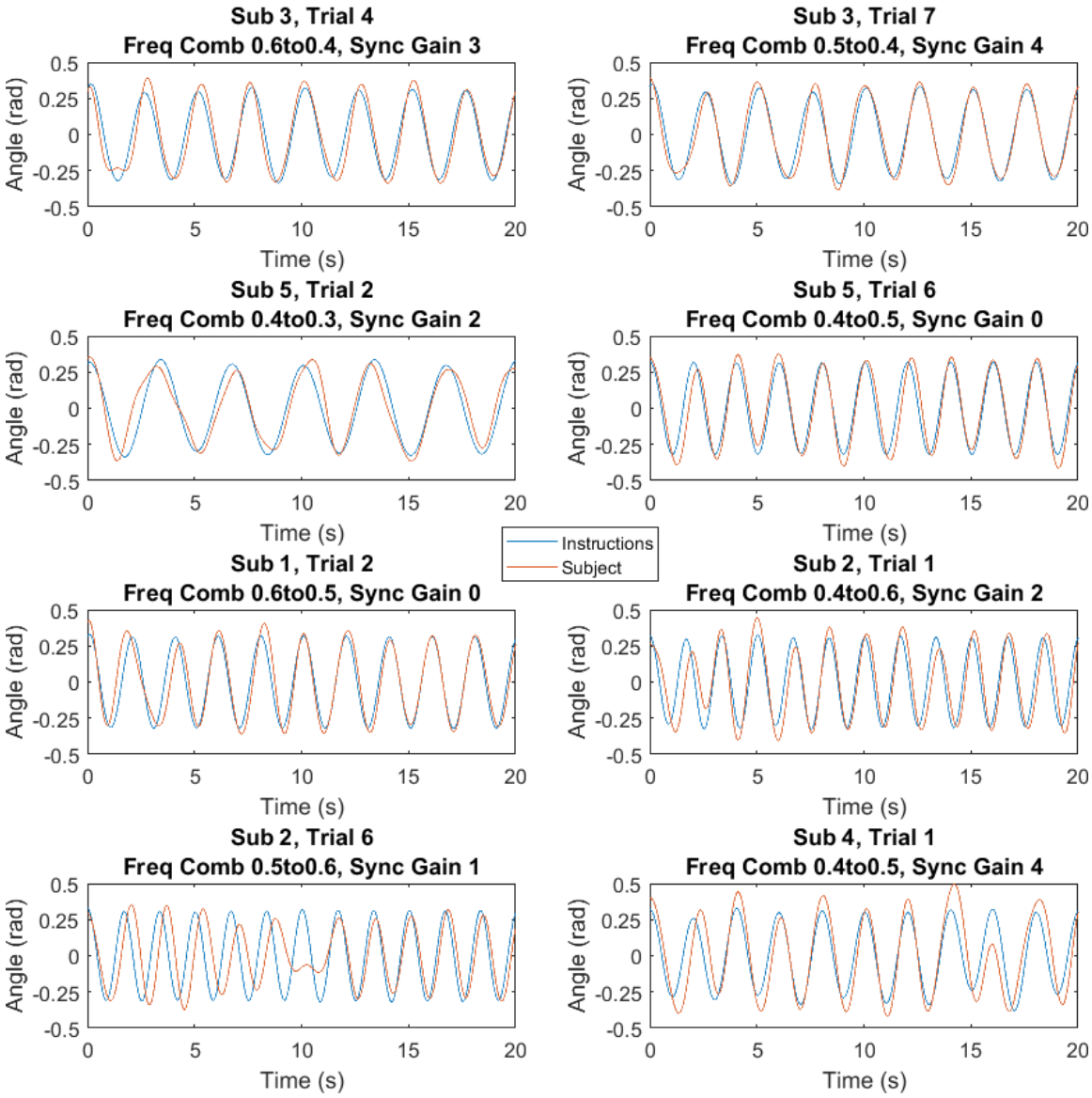



**Figure B.8. Sample data**

## Curriculum Vitae

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Universities attended (with dates and degrees obtained):

Saint Mary's University (Dip Eng in Mechanical Engineering, May 2014)

Dalhousie University (B.Eng in Mechanical Engineering, May 2017)

Publications:

Herod, T. W., Chambers, N. C., & Veres, S. P. (2016). Collagen fibrils in functionally distinct tendons have differing structural responses to tendon rupture and fatigue loading.

Chambers, N. C., Herod, T. W., & Veres, S. P. (2018). Ultrastructure of tendon rupture depends on strain rate and tendon type.

Conference Presentations: None