Inner Ear Driven Balance

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***Abstract* — The ears, integral organs in the human body, play vital roles in hearing and balance. The inner parts of the ears, particularly the semicircular canals (labyrinth) and vestibular system are essential for maintaining body balance. These structures, filled with fluid and lined with hair cells, respond to head movements by shifting the fluid, allowing the brain to sense these movements and control the body to stay balanced. Disturbances or abnormalities in the vestibular system can lead to challenges in maintaining balance, and correct eye position, and may result in vertigo. Understanding the pathology and symptoms associated with inner ear issues provides valuable insights for developing future treatments and care strategies. In this paper, we attempt to model the dynamics of hair cells during head movement. Its mathematical equations were utilized to build a Simulink model, whose output was analyzed.**

***Keywords* — Cochlear hair cells, Stereocilia, Labyrinth, Vestibular System, Simulink**

I. INTRODUCTION

Every human has a basic need for physical activity to thrive physically and mentally in their life[1]. A crucial part of physical activity is balance. The balance enables us to control and to have the right coordination over our body. We can't stand straight or feel stable when we lose a sense of balance. Our central nervous system (CNS) and sensory systems work together to have a proper sense of balance in our body[2]. Sensory systems such as our inner ear, vision, and muscles send a flow of information to our CNS where integration of information happens to control our body, and CNS can control our body based on the information from our sensory system. An issue within one of the sensory systems or CNS still can lead to balance problems.

The auditory and equilibrium functions of our organs, particularly the ears, play integral roles in maintaining vital bodily functions. The inner ear has the cochlea responsible for hearing, while the semicircular canals (labyrinth) and the vestibule, are responsible for the inner ear's role in maintaining balance [3]. There are three different semicircular canals: anterior, lateral, and posterior. The vestibule consists of utricle and saccule. Both structures are filled with fluid and lined with specialized hair cells, called Stereocilia, the labyrinth and vestibule dynamically respond to movement[4]. The labyrinth is responsible for rotary motion, or motion not in straight motion, and the vestibule is responsible for forward/backward and up/down motion. As the body, particularly the head, moves, the fluid within these structures shifts, subsequently affecting the orientation of the hair cells. This interaction enables our brain to perceive movement accurately and exert control over the body to ensure the balance is maintained.

Issues with inner ear conditions lead to the development of multiple diseases such as labyrinthitis and vertigo. Both diseases have symptoms of dizziness and a feeling of off-balance. In England and Wales, the total number of hospital admissions related to the inner-ear issue has increased by 234.8% from 1999 to 2020[5].

In this paper, we aimed to model and simulate the dynamics of the inner ear's hair cell during head movement balance based on relevant physiological details and resources. We designed a closed-loop feedback control system with assumptions and linear time-invariant mathematical equations. SIMULINK is mainly used to simulate a normal model and a disease(vertigo) model of the inner ear hair cell. Their contrasting results from the two models are discussed as well

II. METHOD

A. Key Assumptions

To streamline the modeling and simulation process for the dynamics of hair cells during head movement, several key assumptions have been implemented. The approach embraces a lumped model for hair cells, presuming uniform movement across the cells on each ear side, and attributes structural rigidity to these cells, negating any bending effects. Once the head movement ends, the lumped model of hair cells stays at their final position. In addition, vision is excluded from this simulation of balance, including no light and closed eyes during the simulation. An error in the system comes from signal production within the hair cells, which occurs around 10 microseconds after the signal. Furthermore, the alignment of the head axis is assumed to be

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parallel to the lumped hair cell model, offering a simplified spatial orientation.

These assumptions collectively create a more efficient simulation process, though it is important to acknowledge that certain parts of our assumptions are deliberately omitted for an isolated look into hair cells and balance.

B. Mathematical Equations/Models

The equation below states that the rate of change of the angle θ over time t (angular velocity) is equal to ω(t). In the context of hair cells, θ could represent the displacement of the hair cell, and ω(t) would be its angular velocity.

$$\frac{d\theta}{dt} = \omega(t)$$

This is the rotational form of Newton's second law, where I is the moment of inertia of the hair cell, dω/dt is the angular acceleration, θ(t) is a restoring torque that is proportional to the displacement (potentially due to elastic properties of the hair cell), bω(t) is a damping torque (resistance), and τ(t) is the external torque applied on the hair cell, perhaps due to the movement of fluid in the inner ear or other forces.

$$I\frac{d\omega}{dt} =- a\theta(t) - b\omega(t) + \tau(t)$$

This represents the rate of change of the measured angle θ_{meas}. It's a form of a high-pass filter that would be used to differentiate the actual position from the measured position, factoring in a time constant, t_{meas}. This could be part of a mechanism that senses the position of the hair cell over time.

$$\frac{d\theta_{meas}}{dt} = \frac{1}{t_{meas}}(\theta(t) - \theta_{meas}(t))$$

The equation for our PID controller includes the proportional (P) and derivative (D) terms and omits the integral (I) term (hence a PD controller). Here, K_p is the proportional gain, and K_d is the derivative gain. The PD controller adjusts the torque τ(t) based on the current error e(t), and the rate of change of this error. This could be a model of how the body adjusts the forces on the hair cell in response to sensed imbalances.

$$\tau(t) = K_p e(t) + K_d \frac{d}{dt}e(t)$$

The error e(t) is the difference between the target angle $\theta_{target}(t)$ and the current angle θ(t). This error is what the body tries to minimize to maintain balance.

$$e(t) = (\theta_{target}(t) - \theta(t))$$

C. Constants

The moment of inertia, I, quantifies an object's resistance to changes in its rotational motion. For slender objects like hair cells, the moment of inertia can be estimated using the formula m times r squared, where m is the mass and r is the rotational radius. Based on the provided information, each hair cell has a mass of kg, with an approximate rotational radius of meters. Thus, the moment of inertia for a single hair cell is calculated as:

$$I_{single} = m \times r^2 = 10^{-15}kg \times (10^{-5}m)^2 = 10^{-25}kg \cdot m^2$$

When considering all 10,000 hair cells in the inner ear collectively as a unified mass, the combined moment of inertia is significantly increased. Each hair cell contributes to the total moment of inertia, which aggregates to:

$$I_{total} = 10^4 \times I_{single} = 10^4 \times 10^{-25}kg \cdot m^2 = 10^{-21}kg \cdot m^2$$

The constant a is related to the system's restoring force, which is usually associated with the object's elastic coefficient or a similar force constant; we cannot calculate a directly. The constant b pertains to the system's damping, which is related to the medium's viscosity or internal resistance. Typically, these are determined through experimental data or a detailed system model. In this model, we deduce the possible values of constants a and b based on simulation results. Furthermore, to facilitate calculations in Simulink simulations, we have scaled up the value of I. We believe that scaling all constants of the system proportionally will not affect the accuracy of the actual results.

$$I = 1\,kg * m^2;$$

$$a = 1\,\frac{kg*m}{s^2};$$

$$b = 3\,\frac{kg*m}{s};$$

$$t_{meas} = 10^{-5}s;$$

D. Transfer Function and Laplacian Form

This represents the transfer function of a second-order linear system, where θ(s) is the Laplace transform of the output angle as a function of time, and τ(s) is the Laplace transform of the input torque. The constants b, and a represent the moment of inertia, damping coefficient, and stiffness of the system, respectively. The denominator polynomial characterizes the dynamic behavior of the system, with s being the complex frequency variable in the Laplace domain. The form of this transfer function suggests a system that could be a model for the rotational dynamics of an object, such as the movement of hair cells in response to forces within the inner ear, given their relevance to balance.

$$\frac{\theta(s)}{\tau(s)} = \frac{1}{Is^2+bs+a} \Rightarrow \frac{1}{s^2+3s+1}$$

This represents the open-loop transfer function of a PD controller, where K_p is the proportional gain and K_d is the derivative gain. The term τ_{meas} represents a time constant associated with the measurement delay in the system. This transfer function is designed to modify the behavior of the system by altering the dynamic response to changes in the error between a desired position (setpoint) and the actual position. It is part of a control strategy to achieve desired performance characteristics such as stability, speed of response, and accuracy. The PD controller's effect is to produce an output (torque, in this case) that is fed back to the system to minimize error and achieve the desired control objective. The simplification indicates specific values for the gains and time constant, providing a more direct representation of how the controller will influence the system's behavior.

$$OL(s) = \frac{K_d s+K_p}{(Is^2+bs+a)(1+\tau_{meas})} \Rightarrow \frac{4s+10}{(s^2+3s+1)(1+10^{-5})}$$

E. Implementation of SIMULINK

III. Results

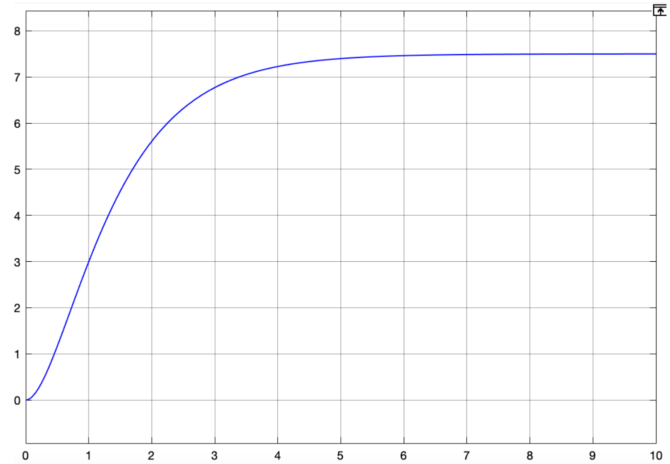

Figure 1: Measured theta in a system with no PD. The target angle was 15 degrees, however, output theta was not achieved, reaching about half the target value.

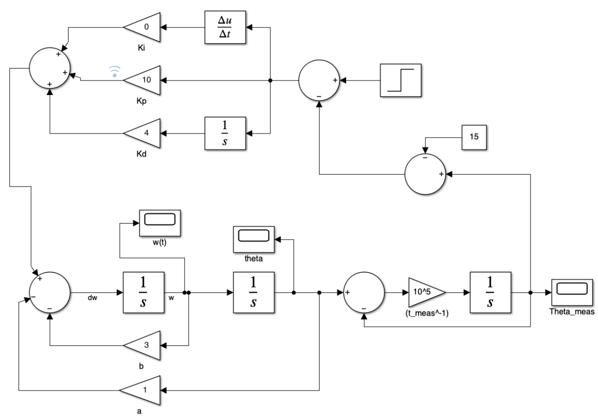

Figure 2: A model with PD in the top left section, hair cell movement in the bottom left section, and time delay and feedback on the right side.

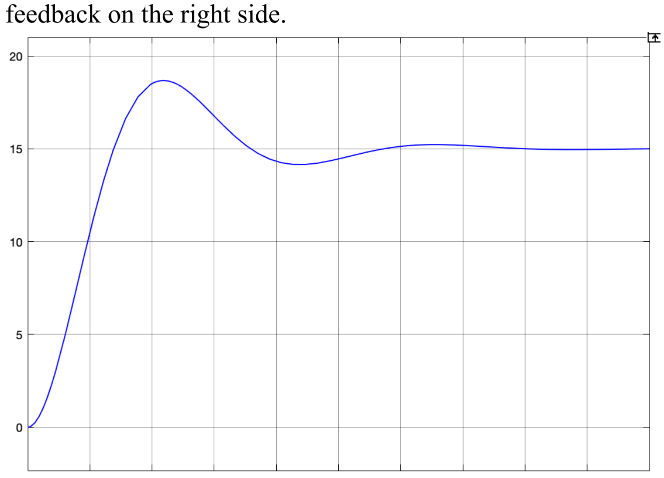

Figure 3: Measured theta graph with the PD control incorporated model; the system reaches the target angle within 4 seconds.

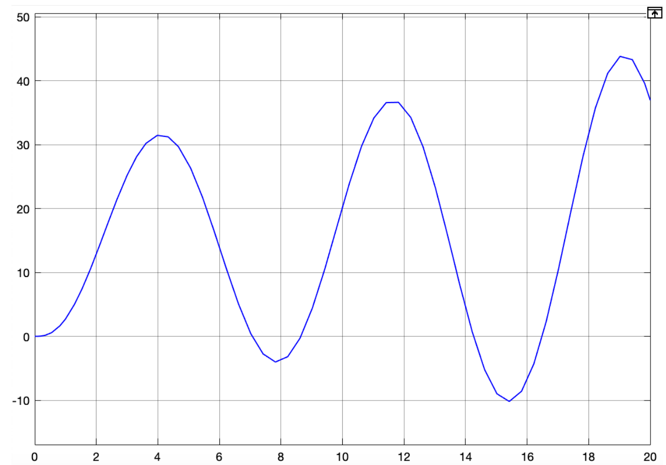

Figure 4: Diseased model with original PD control. Oscillations grow over time due to the large time delay, causing the sensation of vertigo. Attuned model with updated PD control.

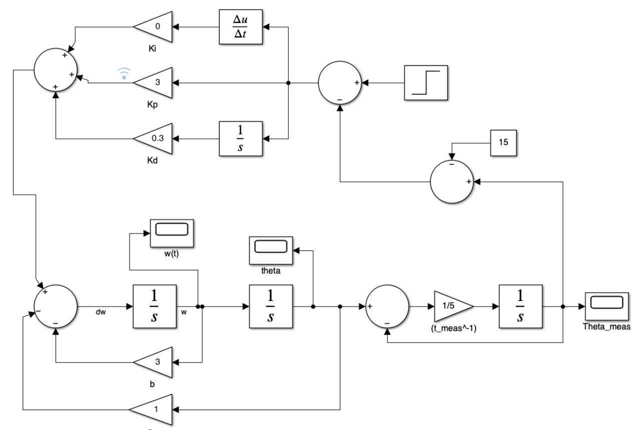

Figure 5: Diseased model with adjusted PD. The PD values decreased which may indicate that when the brain adapts to vertigo, it tones down the input of an ear sending extreme signals.

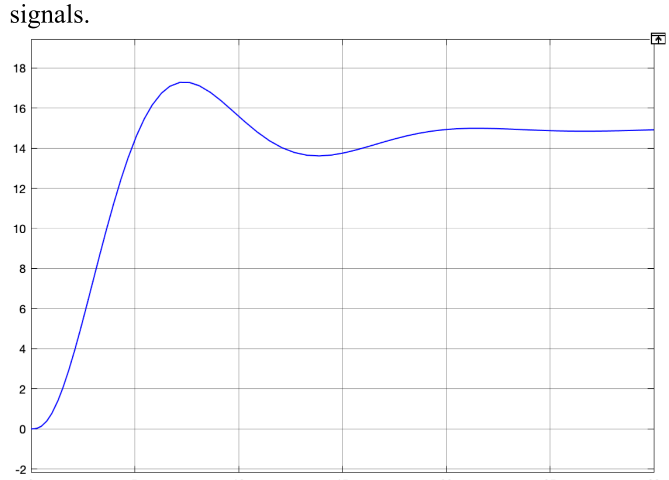

Figure 6: Measured angle in diseased system with updated PD control. The system reaches a target angle of 15 degrees within 25 seconds. This is much longer than the previous system, however, due to the diseased system, a faster time was not possible to achieve.

IV. Discussion

The simulated movement of an inner ear hair cell without a PID controller demonstrated that the current system only reached half the target angle. This necessitated the need for a PD controller to represent our system better. The original PD controller utilized normalized values since most constants found were very small. This PD system was crafted with a Kp of 10 and Kd of 4 which stabilized the system with a -90 phase margin.

The change in time delay was increased to demonstrate a diseased system where the hair cells are damaged causing either weaker signals or lack of signal production. The increase from 10 microseconds to 5 seconds was chosen since it represents a possible median of time delays in signal production from hair cells in individuals with inner ear disorders, however, there is no literature on the actual time delays in signal production for these conditions. As shown in the previous figures, any notable increase in time delay would have caused the system to become unstable.

The new PD controller utilized a smaller Kp of 3 and Kd of 0.3, which could represent the need to quiet faulty data from a damaged ear. If the brain took the input from a diseased ear without adjusting, it would be susceptible to constant vertigo and other balance issues.

V. Conclusion

The model with the PD controller successfully simulates the movement of inner ear hair cells and correlates their deflection with head position. The conversion of this system to a diseased model does give a quantitative representation of vertigo as well. However, the model is limited due to the many unknown quantities of the subject. Since the hair cells are very small and their physical properties are not well studied, it was very difficult to format the equations and find solid results. This forced the model to be very simple since adding complex feedback loops could not be supported. Research into the dynamics of these hair cells could heavily improve our model. Another area that can be expanded on would be incorporating two separate systems for the vestibule and semicircular canals (labyrinth) which sense forward/backward movement and rotational movement, respectively. Having two separate systems would allow the system to track movement in 3D space, giving a more applicable system that could stimulate hair cell sensation and where issues like vertigo can crop up. These systems could also account for movement of fluid in the inner ear, which would make the system more complex but accurate. Other inputs that do influence balance like vision and proprioception would be helpful in regulating the ear system, however, these inputs are very difficult to implement due to their abstract units. Our model can serve as a base to start designing solutions to inner ear issues related to sensory cells and could be applied to show a mathematical representation of balance issues related with the inner ears.

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