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UTILIZING BANDWIDTH TO QUANTIFY HUMAN TORSO MOTOR CONTROL CAPABILITY

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INTRODUCTION

Although the methods used to quantify bandwidth for a simple engineering system are well established [1], the application of bandwidth to complex systems like a human is more challenging. A simple explanation is that higher bandwidth controllers are able to respond to systems that change more quickly. Many different factors influence the human neuromuscular control system, and its response to external inputs. Some of these factors include age, physical health, neurological condition, external environment and body mass distribution. The human neuromuscular control system is further complicated by the fact that it may have different responses to the same input signal. Since there are a large number of confounding factors and since they are inherently coupled with the system, a large amount of experimental data is needed to understand the influence of each factor on bandwidth.

In order to understand the complexity of human motor control, some researchers have used mathematical models to study torso movement. Typically, input commands are presented to the system and the motor control moves the body to track these movement commands.

The ultimate goal of this research is to develop new diagnostic methods that can be used by medical professionals to assess the degree of neuromuscular disease and evaluate the effectiveness of treatment. In this research, methods are developed to calculate the bandwidth of the neuromuscular control system which may be used as a measurable variable to quantify neuromuscular controller capability.

METHODS

The upper body (including the head, arms and torso) was represented as an ellipsoid (Figure 1). The height and width of the ellipsoid were selected based on the values of typical human subjects. A simulated motor was connected to the upper body to control the rotation of the ellipsoid.

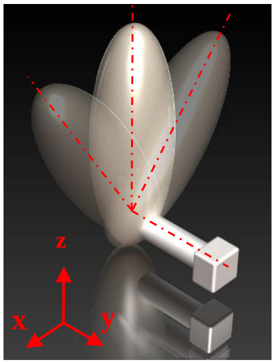

Figure 1: Representation of upper body and controller.

The ellipsoid representing the upper body was assigned a moment fo inertia of *I*. Torso stiffness due to spinal ligaments, and passive muscle tone of the abdominal and back muscles were assigned a stifness value of *k*. Viscous damping of the torso during movment was assigned a value of *b*.

A second order differential equation was used to describe the dynamics of the system. A general rotational dynamic system can be represented by

$$I\ddot{\theta} + b\dot{\theta} + k\theta = u(t) \qquad (1)$$

where θ is the angular position about the lumbar spine and *u(t)* is the torque applied to the system. In this model, *u(t)* consists of two components.

The first is torque generated by active contractions of the abdominal and back muscles. This torque is controlled by the human motor controller in the brain. The second component is the effect of gravity on the system. A mathematical representation of these torques is given by

$$u(t) = G(\theta_d - \theta) + mgl\,sin(\theta) \quad (2)$$

where G is the gain of the controller. The paremeters m, g and l are the mass of upper body, acceleration due to gravity and the length of the segment to the center of mass, respectively. θ_d is the desired angular position. Using Euler's approximation, the mathematical representation of the system for small angles is

$$I\ddot{\theta} + b\dot{\theta} + \theta(k - mgl) = G(\theta_d - \theta) \quad (3)$$

Representative values for the unknown parameters I, b, k, m, l, and G were selected based on typical human values [2]. These unknown model parameters are stored in the vector $\alpha = (I, b, k, m, l, G)$. The mathematical model was run to generate a system response to a step function. Six step functions were evaluated (±3º, ±2º, ±1º). White Gaussian noise was added to the response curves and the curves were filtered with a band pass filter over the range of 0.01 to 10 Hz. These curves are show as the black lines in Figure 2. Each curve is an average of three separate simulation runs. The curves serve as a surrogate for data that will be collected in future human subject experiments.

Our method uses optimization to determine the unknown parameters in Equation 3. The response of the model is compared to the simulated experimental data. Optimization is carried out by minimizing the cost function shown below,

$$S = \sum(\theta_{s,i} - \theta_{m,i}(\alpha))^2 \quad (4)$$

The values for the parameters in vector α can be determined from the solution of the optimization problem. Variable $\theta_{s,i}$ is the angular position of the simulated experimental data at the i^{th} time step (black line in Figure 2). Variable $\theta_{m,i}(\alpha)$ is the angular position of the model response given parameters α. The optimized curves are shown in Figure 2 (colored lines).

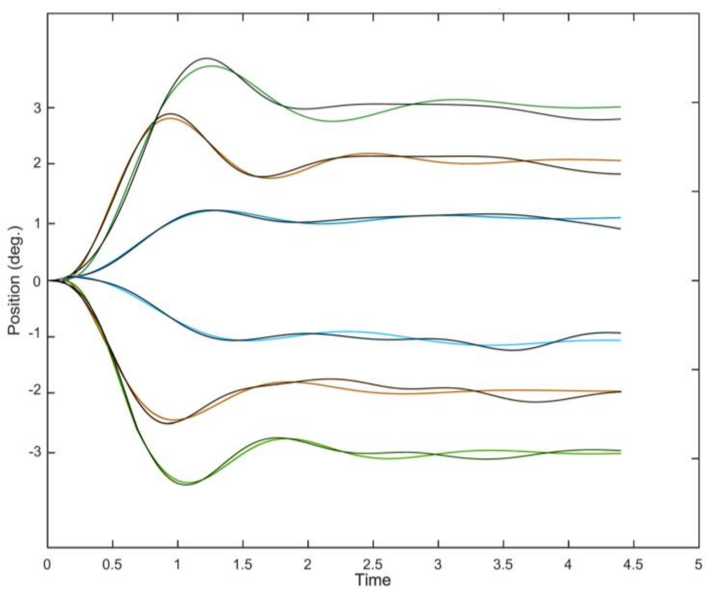

Figure 2: Generated data (black) vs. model (colored).

Now that we have the system response to a step function, we can calculate the bandwidth. A Fourier transform is applied to Equation 3 yielding,

$$\theta_{(j\omega)}(I(j\omega)^2 + b(j\omega) + k - mgl) = G\left(\frac{\theta_d}{j\omega} - \theta_{(j\omega)}\right) \quad (5)$$

where, $\theta_{(j\omega)}$ is the angular position in frequency domain, j is an imaginary number equal to $\sqrt{-1}$, and ω is the frequency. The transfer function, $\theta_{(j\omega)}/\theta_d$ = $Tr_{(j\omega)}$ = Re(jω) + Im(jω), is determined by manipulating Equation 5. The transfer function has both real and imaginary components. The magnitude response of the Bode plot is given by $mag = \sqrt{Re^2 + Im^2}$, and plotted with respect to frequency (Figure 3).

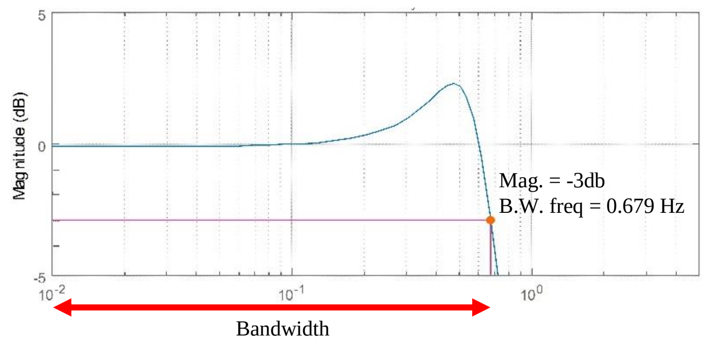

Figure 3: Typical Magnitude Response of Bode Plot.

The bandwidth of a system is defined as the range of frequencies over which the magnitude response of the transfer function drops by 3 dB. This drop in amplitude corresponded to a 50% decrease in power.

RESULTS

The bandwidth frequency was calculated for each of the six step functions and is listed in Table 1. The magnitudes of the step functions were defined as small (1º), medium (2º), and large (3º).

Table 1: Bandwidth frequency (Hz) for different desired positions

		Small	Medium	Large
Direction	Flexion	***0.58***	***0.84***	***0.80***
	Extension	***0.66***	***0.91***	***0.68***

DISCUSSION

Overall, the magnitude of bandwidth frequencies was seen to be fairly consistent, ranging from 0.58 to 0.91 Hz. We see different bandwidth for different input magnitudes and direction. Hence, bandwidth must be a function of magnitude and direction of the motion.

The results of this analysis are promising. The analysis method was applied to surrogate data in the form of a time series. In the future, this method will be used to analyze time series data collected experimentally. This will enable the bandwidth of the human neuromuscular control system to be determined.

REFERENCES

1. Phillips, C., & Harbor, R., 2000. Feedback control systems (4th ed.). Upper Saddle River, N.J.: Prentice Hall
2. Reeves, N., Cholewicki, J., Narendra, K., 2009. Effects of reflex delays on postural control during unstable seated balance. J.Biomech.42,164–170.