2-1-2004

Sensitivity Analysis of Hill Muscle Parameters

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1. ABSTRACT

A computational, rigid body model of a 50th percentile male head and neck utilizing15 Hill Muscle pairs is used to study the sensitivity of Hill Muscle Model parameters. A 15g linear acceleration is applied within the transverse plane at the lowest vertebral level of the neck (T1). The resultant linear acceleration of the head is analyzed. In comparing the resultant linear acceleration of the head, the timing of the acceleration response is minimally affected. The peak accelerations did change, and in the case of varying muscle activation, the peak acceleration changed significantly, 36%. Each of the other parameter variations affected the peak acceleration of the head by less than 5%. Overall, the muscle activation parameter has the most significant influence on the response of the system.

2. SYSTEM OVERVIEW

There are numerous parameters within the computational Hill Muscle Model. Some values can be measured but others must be chosen from recommended values determined from various experimental studies. In order to obtain optimal force characteristics when applying Hill Muscle Models, it is imperative to understand the sensitivity of each of the underlying parameters. This paper addresses the study of system response to variations within the Hill Muscle parameters within a computational, rigid body model of a 50th percentile male head and neck utilizing15 Hill Muscle pairs.

In this study, a lumped parameter head and neck computational model developed by Brelin-Fornari (1998) is utilized. The model was developed using MADYMO™, a commercially available, rigid body/finite element, dynamic analysis software package. The three dimensional, sagittal plane symmetrical model of the head and neck consists of ten rigid body masses. These masses are joined by forces, modeled as linear viscoelastic intervertebral joints allowing full six degrees of freedom relative motion between adjoining masses, and fifteen symmetrical pairs of active muscle elements (Brelin-Fornari 1998).

The 15 pairs of active muscles were modeled using Hill's methodology. A typical Hill muscle model consists of an elastic element (SE), in series with a contractile element

Keywords: muscle model, sensitivity study, Hill model, neck model

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(CE), in parallel with a spring passive element (PE) (Figure 1 A and B). In this study, the effect of the series elastic element was deemed negligible since the effect of that element becomes unimportant in the presence of large changes in muscle length (Close 1972). Therefore, the total force generated by the muscle at any given time is given by:

$$F = F_{ce} + F_{pe} \tag{1}$$

Where F_{ce} is the force output of the contractile element and F_{pe} is the force output of the passive element.

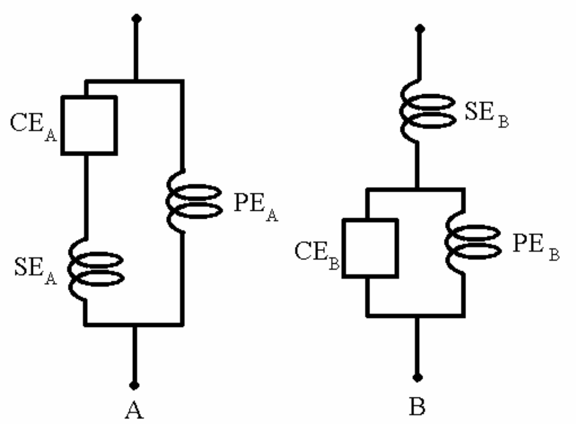

Figure 1 A and B: Classic structures for the Hill muscle model

In the Hill model, the action of the contractile element represents the force generated by the cross-bridges of the muscle induced by the chemical reaction within the muscle. The input to the contractile element is a neural impulse. The output is a force (F_{ce}), which is a function of the normalized muscle length (l_r) and rate of change of its length (or velocity, v_r), activation level (A), and the maximum force available at maximum activation (F_{max}) (Winters 1990). This can be written as:

$$F_{ce} = AF_{max} f_h(v_r) f_l(l_r) \tag{2}$$

Hill (1949) defined the "active state" of a muscle as the state where the CE generates tension, without lengthening or shortening, after the beginning of excitation. The value of activation, A, ranges from 0% (approximately .5% for muscles at rest) to 100% (full activation). If activation is zero, only the passive element is generating force ($F_{ce} = 0$).

The maximum force generated during a shortening contraction, F_{max}, is assumed to be independent of the fiber composition (slow or fast fibers) and dependent only on the physiological cross-sectional area (A_{pcs}) (Winters 1985). This relationship can be expressed as:

$$F_{max} = \sigma_p A_{pcs} \tag{3}$$

Where σ_p is the peak muscle stress and A_{pcs} is the physiological cross-sectional area. In this study, σ_p was assumed to be constant at 0.4 MPa, which is the value reported by Yamada (1970) for the sternocleidomastoid (a dominant neck muscle) and corrected for living tissue (ultimate tensile strength postmortem, 19 g/mm^2, is 50% of that just after death (Yamada 1970). The physiological cross-sectional area represents the sum of the cross-sectional areas of all the muscle fibers within a muscle, parallel to the force generating axis of the muscle. Typically, the muscle fibers are not oriented along the force generating axis but at some angle with respect to the axis. This angle is referred to as the pennation angle θ. If the pennation angle is not zero, the physiological cross-sectional area is defined as:

$$A_{pcs} = \frac{m\cos\theta}{\rho l_f} \tag{4}$$

where m is muscle mass (in grams), ρ is the muscle density (1.056 grams/cm^3 for mammalian muscle (Lieber, 1992)), and l_f is the fiber length (in cm) (Winters 1985). Myers (1998) reported A_{pcs} for several neck muscles.

The force velocity (F-v) relation for shortening muscle ($-1 < v_r \leq 0$), was identified by Hill (1938) as the hyperbolic relationship:

$$(F + a)\,(v + b) = (F_{max} + a)\,b \tag{5}$$

Equation (5) can be solved for F/F_{max} therefore defining the force-velocity function as:

$$f_h(v_r) = \frac{F}{F_{max}} = \frac{(1+v_r)}{1-\dfrac{v_r}{a_f}} \tag{6}$$

where v_r is the normalize muscle velocity and a_f is the hyperbolic shape factor. Since neck muscle fibers have evenly distributed fast and slow fiber compositions (Winters and Woo 1990), a_f has been reported as 0.25 (Winters 1990) and v_{max}, the normalization factor for v_r, was calculated as six times the reference length, l_{ref}, per second, for each muscle (Winters and Stark 1985). The variable v_{max} can range in value from two times the reference length per second for slow fibers to ten times the reference length per second for fast fibers (Winters and Stark 1985).

The force-length relation (F-l) is based on the observation that the amount of tension developed by a muscle fiber can be altered by changing the length of the fiber prior to contraction. A wide variety of empirical fits have been utilized to describe the F-l relation. A popular form of the parabolic curve is given as:

$$f(l_r) = \exp\left[\frac{-(l_r-1)^2}{S_k}\right] \tag{7}$$

where l is the muscle length, l_{ref} is the optimal length, l_r is the normalized length $l\,/\,l_{ref}$, and S_k is the “shape” function (width) of the parabolic curve. Commonly used values of S_k range between .40 and .68 (Winters and Stark 1985). The optimal length (l_{ref}) for the neck muscles can be calculated from the linear relation:

$$l_{ref} = l_{rest}\left(\frac{s_{ref}}{s_{rest}}\right) \tag{8}$$

Rack and Westbury (1969) determined optimal sarcomere length (s_{ref}) as 2.8 - 3.0 µm. The value of l_{rest} is the length of the muscle *in situ*, when the body is at rest, and s_{rest}, for each muscle was reported by Myers (1998). The measurement of s_{rest} was performed using phase contrast and laser diffraction methods with full DIC, Nomarski optics. The sarcomere length was determined to an accuracy of ± .025 µm. Complete sarcomere measurement information can be found in Myers (1998).

A passive muscle is a muscle without neural input. The characteristics of the passive muscles can be found in Brelin-Fornari (1998).

An acceleration pulse (Figure 2) was applied at the T1 level of the head/neck model. Gravitational and muscle pretensioning forces where also applied.

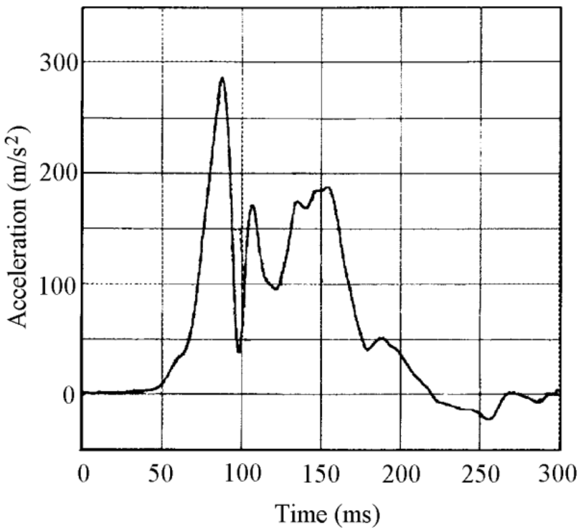

Figure 2: Acceleration pulse applied to the T1 level of the Head/Neck Model

3. RESULTS

The resultant head acceleration was recorded for each of the variations of the Hill Muscle parameters. In comparing each of the iterations, the timing of the acceleration response is minimally affected. The peak accelerations did change, and in the case of varying muscle activation, the peak acceleration changed significantly, 36% (Figure 3).

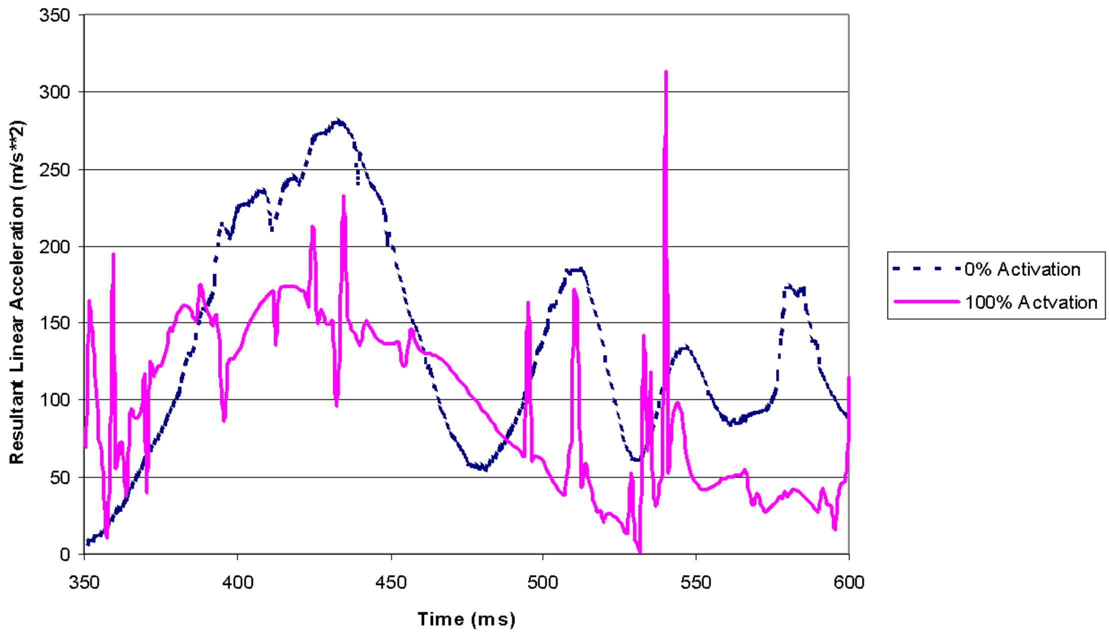

Figure 3: Comparison of Resultant Linear Acceleration for 0 and 100% muscle Activation

Table 1 lists the complete results of the muscle parameter sensitivity study. It is clear from this analysis, that the muscle activation parameter (*A*) has the greatest influence on the kinematics of the head cg. A 36% change in the peak resultant linear acceleration was calculated from the minimum and maximum values of activation while each of the other parameters changed the kinematics by less than 5%.

Table 1: Muscle Parameter Sensitivity Analysis

Muscle Parameter	**Parameter Values**	**Peak Resultant Linear Acceleration* (m/s^2)**	**Change from Min to Max of Parameter Range (%)**
Muscle Activation, *A*	0%	280	36.0
	100%	180	
Maximum shortening Velocity, v_{max}	$2 * l_{ref}$	341	3.6
	$4 * l_{ref}$	311	
	$6 * l_{ref}$	309	
	$8 * l_{ref}$	308	
	$10 * l_{ref}$	329	
Ultimate Strength as a function of maximum force produced, F_{max}	1.1 F_{max}	323	4.8
	1.3 F_{max}	310	
	1.5 F_{max}	306	
	1.8 F_{max}	308	
Optimal Sarcomere Length, S_{ref}	2.8 μm	309	0.6
	2.9 μm	308	
	3.0 μm	312	
Shape Function (width), S_k	0.4	308	2.0
	0.54	313	
	0.68	315	

* Note - The peak resultant linear acceleration of the head cg occurs at approximately 100 milliseconds after the onset of the sled pulse.

4. CONCLUSIONS

The Hill Muscle Model consists of an array of parameters. Some of the parameters can be measured for a specific muscle, such as sarcomere length. Other variables must be estimated using previously conducted laboratory experiments. For the estimated values, a range has been reported in literature. Using the reported values, a sensitivity analysis is performed on a computational, rigid body model of the head/neck complex to determine which of the variables has the most profound effect on the outcome of the analysis.

In comparing the resultant linear acceleration of the head, the timing of the acceleration response is minimally affected. The peak accelerations did change, and in the case of varying muscle activation, the peak acceleration changed significantly, 36%. Each of the other parameter variations affected the peak acceleration of the head by less than 5%. Overall, the muscle activation parameter has the most significant influence on the response of the system.

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