Introduction

Chapter 1

Systems Perspective

2. Terminology from a Movement Science Perspective

1. Movement Science: Foundations and Terminology

The purpose of this chapter is to provide a conceptual framework and nomenclature for the description and analysis of movement. This chapter introduces key concepts and terminology used in the field of movement science, enabling a comprehensive understanding of the movement process.

Section 7

Modern movement science (section 7) is a field that focuses on the study of movement patterns, kinematics, and dynamics. It integrates knowledge from various disciplines, including bioengineering, biomechanics, and neurophysiology, to provide a holistic understanding of human movement.

Section 8

Theories and models of movement (section 8) are developed to explain the observed movement patterns and their underlying mechanisms. These theories help in predicting movement behavior and identifying factors that influence movement.

Section 9

Instrumentation and measurement of movement (section 9) is crucial for the quantification of movement parameters. This chapter discusses various methods and tools used in measuring movement, including the use of motion capture systems, force plates, and electromyography (EMG).

Section 10

The interdependence of movement (section 10) highlights the complex relationship between different movement components. Understanding these interdependencies is essential for the development of effective movement strategies and interventions.

Section 11

Movement control and coordination (section 11) examines how the nervous system controls and coordinates movements. This section focuses on the neural mechanisms that underlie movement control and the factors that influence movement coordination.

Section 12

Movement and health (section 12) explores the relationship between movement and health outcomes. This chapter discusses how movement plays a crucial role in maintaining physical health and well-being.

Section 13

Movement and performance (section 13) investigates the role of movement in athletic and non-athletic performance. This section examines how movement can be optimized for better performance in various contexts.

Section 14

Movement and society (section 14) considers the impact of movement on society and vice versa. This chapter explores how societal factors influence movement and how movement shapes societal structures.
2.2 Muscular Mechanisms

Muscle

\[ \text{Muscle} \]

Proximal of origin

Distal of insertion

2.2.1 Tension-Displacement Loops

A tension-displacement loop is a curve that represents the relationship between the tension produced by a muscle and the displacement of its insertion point. The loop is typically measured in an isolated muscle preparation and is used to determine the force-length relationship of the muscle. The loop is characterized by an initial linear portion, where tension increases linearly with displacement, followed by a non-linear region where tension increases at a slower rate.

2.2.2 Contractile Elements

The contractile elements of the muscle are responsible for the production of force. These elements include the myofibrils, which are the contractile units of the muscle, and the sarcoplasmic reticulum, which stores calcium ions that are necessary for muscle contraction.

2.2.3 Neuromuscular Transmission

Neuromuscular transmission is the process by which electrical signals from the motor neurons are converted into mechanical signals that cause muscle contraction. The process involves the release of neurotransmitters from the presynaptic terminal, which stimulate the postsynaptic membrane to depolarize, leading to the generation of action potentials and muscle contraction.

2.2.4 Muscle Fiber Types

Muscle fibers are classified into different types based on their contractile properties and oxidative capacity. Type I fibers are slow-twitch, oxidative fibers that are primarily involved in endurance activities. Type II fibers are fast-twitch fibers that can be further subdivided into Type IIa (fast-twitch, oxidative-glycolytic) and Type IIb (fast-twitch, glycolytic) fibers, which are primarily involved in explosive activities.

2.2.5 Muscle Fiber Recruitment

Muscle fibers are recruited in a specific sequence during exercise, with Type I fibers being recruited first and Type II fibers recruited later as the intensity of the exercise increases.

2.2.6 Muscle Fatigue

Muscle fatigue is a decrease in muscle strength and endurance that occurs during prolonged exercise. It is caused by the depletion of energy stores and the accumulation of metabolites in the muscle fibers. Techniques such as active and passive stretching, cooling, and the use of oxygen supplementation can help delay the onset of fatigue.

2.2.7 Muscle Adaptation

Muscle adaptation refers to the changes that occur in muscle structure and function in response to exercise. These changes include increases in muscle mass (hypertrophy), increases in muscle strength, and improvements in muscle endurance.

2.2.8 Muscle Performance

Muscle performance is a measure of the ability of a muscle to produce force and power. It is influenced by factors such as muscle fiber type, muscle architecture, and the nervous system.
1. The term "compressed air" is often used in conjunction with the pressure of the flow of air. However, it is preferred to use "elastically compressed air" to avoid confusion with the "compressed air" used in pneumatic systems.

2. The process of compression is essential to the operation of many pneumatic systems. It involves the conversion of kinetic energy into potential energy, which is then used to power the system.

3. The diagram illustrates the relationship between the pressure and volume of the air. As the pressure increases, the volume decreases, and vice versa.

4. The equation for calculating the pressure of the air is given by $P = \frac{F}{A}$, where $P$ is the pressure, $F$ is the force, and $A$ is the area.

5. The diagram shows the different stages of compression, including the intake, compression, and delivery stages.

6. The intake stage involves the intake of air into the pneumatic system. The compression stage involves the conversion of kinetic energy into potential energy, while the delivery stage involves the delivery of the compressed air to the desired location.

7. The efficiency of the compression process is determined by the ratio of the final pressure to the initial pressure. A higher ratio indicates a more efficient compression process.

8. The diagram also shows the different components of a typical pneumatic system, including the air compressor, the air dryer, and the air receiver.

9. The air compressor is responsible for compressing the air, while the air dryer removes any moisture from the compressed air. The air receiver stores the compressed air for future use.

10. The diagram also shows the different types of pneumatic systems, including the compressor, the dryer, and the receiver.

11. The efficiency of the compression process is determined by the ratio of the final pressure to the initial pressure. A higher ratio indicates a more efficient compression process.

12. The diagram shows the different types of pneumatic systems, including the compressor, the dryer, and the receiver.
3.1 Muscle Action

Mechanokinetic Mechanics

3.3 More Terminology

Hyperbolic Functions

2.6 HNN Model vs. Standard HNN Model

The hyperbolic functions (HNN) model is an alternative to the standard HNN model, which uses hyperbolic functions to represent the connection weights. The hyperbolic functions have the advantage of being continuously differentiable, which makes them more suitable for neural network implementations. The hyperbolic tangent (tanh) function is often used in neural networks as an activation function because it maps the input values to a range of -1 to 1, which can help in stabilizing the training process. In contrast, the standard HNN model uses a step function, which can lead to more abrupt changes in the output. The use of hyperbolic functions in the HNN model allows for smoother transitions and can potentially lead to better performance in certain tasks.
Section 3

In 1990, we will develop a new computer display technology that can be used to enhance the performance of existing systems. This technology involves the use of advanced materials and processes that promise to revolutionize the way we interact with our computer systems. The benefits of this new technology include increased efficiency, reduced costs, and improved user experience.

We have already begun preliminary studies to determine the feasibility of this technology. These studies have shown promising results, and we believe that we are on the verge of making significant breakthroughs in the field.

The potential applications of this technology are vast, ranging from personal computers to large-scale systems used in industry and government. We are excited about the possibilities that this new technology offers and are committed to advancing it as quickly as possible.

References


Figure 1: Schematic diagram of the new display technology.
$$\frac{P}{Q} \leq -\ln\left(\frac{N}{Q}\right)$$

**Example Text:**

- **Equation:** $\frac{P}{Q} \leq -\ln\left(\frac{N}{Q}\right)$
  
- **Description:** This equation represents a fundamental concept in information theory, specifically the relationship between probability and information entropy. It is used to quantify the amount of information contained in a message.

- **Diagram:** The diagram illustrates a flowchart or process diagram, likely related to the concept of information flow or decision-making processes.

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**Additional Information:**

- The text seems to be discussing the principles of information theory, focusing on the concepts of probability, entropy, and their applications in various contexts.
- The presence of diagrams and equations suggests a detailed exploration of these topics, possibly within the context of computer science, data analysis, or electrical engineering.

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**Note:** The text is cut off at the edges, and the full context is not visible. The content provided is a faithful representation of the visible portion of the text.
3.6 Dynamic Functions of Motion

3.6.1 Linear Motion

The position of a moving object can be described by its position vector, which is a function of time. The position vector, \( \mathbf{r} \), at any time \( t \), is given by

\[
\mathbf{r}(t) = \mathbf{r}_0 + \mathbf{v}t + \frac{1}{2} \mathbf{a}t^2,
\]

where \( \mathbf{r}_0 \) is the initial position, \( \mathbf{v} \) is the initial velocity, and \( \mathbf{a} \) is the acceleration.

3.6.2 Acceleration

The acceleration vector, \( \mathbf{a} \), is the derivative of the velocity vector with respect to time. If the acceleration is constant, then the acceleration vector is a constant vector.

3.6.3 Force and Motion

According to Newton's Second Law of Motion, the net force acting on an object is equal to the mass of the object multiplied by its acceleration. This can be expressed as

\[
\mathbf{F} = m \mathbf{a},
\]

where \( \mathbf{F} \) is the net force, \( m \) is the mass of the object, and \( \mathbf{a} \) is the acceleration.

3.6.4 Motion under Constraints

When an object moves under constraints, its motion is limited to certain paths or regions. The constraints can be represented by equations that describe the boundaries or limitations.

3.6.5 Motion in Non-Inertial Frames

When an object moves in a non-inertial frame, the equation of motion is not the same as in an inertial frame. The centrifugal force and Coriolis force are additional terms that must be considered.

3.6.6 Time-Dependent Motion

When the position, velocity, or acceleration of an object depends on time, the motion is considered time-dependent. The equations of motion are then time-dependent differential equations.

3.6.7 Examples of Motion

Examples of motion include simple harmonic motion, projectile motion, and circular motion. Each type of motion has its own set of equations and methods for solving the motion equations.
4.3 Types of Physical Evidence

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In the broadest sense of the term, evidence can be classified into two categories: tangible and intangible. Tangible evidence includes physical objects, documents, and other items that can be touched or observed. Intangible evidence, on the other hand, consists of information or data that cannot be touched or observed, such as photographic evidence, audio recordings, and written statements.

4.4 Information Processing

The process of information processing involves the transformation of raw data into useful information. This includes the collection, organization, and analysis of data to extract meaningful insights. Information processing is crucial in fields such as business, science, and healthcare, where data-driven decisions are essential.

4.5 Memory and Function

Memory refers to the ability to store and recall information. It is a complex cognitive process that involves the encoding, storage, and retrieval of information. Memory is essential for learning, problem-solving, and decision-making. The human memory system is divided into three main components: sensory memory, short-term memory, and long-term memory.

4.6 Neuron Synapses

Neurons are the basic building blocks of the nervous system. They communicate with each other through chemical signals called neurotransmitters. Neuron synapses are the points of contact where information is exchanged between neurons. The strength of these synapses, known as synaptic plasticity, can change with experience and learning, allowing the brain to adapt and learn over time.
I, the undersigned, do solemnly swear that I will faithfully execute the office of the...
The process of information is not voluntary and can be influenced by various factors such as emotions, perceptions, and context. The information path is often influenced by the receiver's prior knowledge and experiences. The旅游 is also determined by the receiver's attitude towards the information. The concept of information processing is crucial in understanding how the human brain processes and organizes information. The information processing model is a framework that describes how information is received, stored, and retrieved. The model consists of four stages: input, storage, retrieval, and output. The input stage involves the reception of information from the environment. The storage stage involves the encoding and storage of information in the memory. The retrieval stage involves the retrieval of stored information from memory. The output stage involves the use of the retrieved information to make decisions or take actions. The information processing model is a useful tool for understanding the human brain's information processing capabilities.
1. Termination Conditions of Momentum Control

2. Issues are Alive

3. Properties and Simulation of Induction

4.地区有温超、间

5. The moment is epochable of momentum control
Fundamental Muscle Mechanics

Chapter 1
Fundamental Music Mechanics

Isolated Muscle Preparation

1. Place the muscle in a stream of water. Do not allow the muscle to come into contact with the tube walls.
2. Insert the muscle into a lather of water. The lather should be a gentle stream of water. Do not allow the muscle to come into contact with the tube walls.
3. Rinse the muscle in a gentle stream of water. Do not allow the muscle to come into contact with the tube walls.
4. Place the muscle into a stream of water. Do not allow the muscle to come into contact with the tube walls.

The muscle is now ready for further experiments.
Further increases in frequency produce no further increases in force. However, if the frequency is increased past a certain point, the muscle fibers begin to fatigue, reducing the overall force output. (6.2.4) When the frequency is increased, the mean force rises and the force-time relationship is given by the function introduced in Section 6.2.

When a force-time relationship is given, the force is described using a mathematical expression that relates the force to time. The relationship is often expressed as a function of time, where the force is a function of time. The relationship can be expressed using a variety of mathematical functions, including linear, quadratic, or exponential functions.

The force-time relationship is important for understanding how muscles respond to different stimulation patterns. For example, if a muscle is stimulated at a high frequency, it will produce a higher force than if it is stimulated at a lower frequency. This is because the muscle fibers are able to generate more force when they are stimulated more frequently.

The force-time relationship can also be used to predict the behavior of muscles in different situations. For example, if a muscle is being used in a repetitive task, such as lifting weights, the force-time relationship can be used to predict the peak force that the muscle will be able to generate, and how long it will be able to sustain that force.

Mechanical Events: Twitch and Tetanus

Mechanical events are the changes that occur in muscle fibers when they are stimulated. When a muscle fiber is stimulated, it produces a mechanical event known as a twitch. A twitch is a brief, rapid contraction of the muscle fiber. This contraction is caused by the release of calcium ions from the sarcoplasmic reticulum, which allows the myofilaments of the muscle fiber to slide past each other, producing a force.

When a muscle fiber is stimulated repeatedly, it produces a series of twitches, known as a tetanus. A tetanus is a continuous series of contractions that occur when a muscle fiber is stimulated at a high frequency. This allows the muscle fiber to maintain a high level of force over a long period of time.

The tetanus is important for understanding how muscles are able to generate and sustain force. By studying the tetanus, researchers can learn about the mechanisms that allow muscles to generate force and sustain it for long periods of time. This information is important for designing exercises and training programs that can help improve muscle function.
Cross-sectional area in cm^2, the elastic modulus of steel is 1.0 GPa/cm^2. 

The stress in the cross-sectional area is the force per unit area. 

The stress in the cross-sectional area is known as the load acting on the cross-sectional area. 

The stress is given by the formula: 

\[ \sigma = \frac{F}{A} \] 

where \( F \) is the force applied and \( A \) is the cross-sectional area.

The stress is measured in units of MPa (megapascals) or psi (pounds per square inch).

The strain is defined as the change in length divided by the original length: 

\[ \varepsilon = \frac{\Delta L}{L_0} \] 

where \( \Delta L \) is the change in length and \( L_0 \) is the original length.

The strain is measured in units of percent (%).
Quick-Reference Experiments

Tension of muscles, such as the long, thin muscles of the leg, are less likely to have a local maximum than parallel-fibered muscles. The proportion of concentric fiber types in the muscle is a major factor in determining the level of tension that can be developed. A hypertrophied muscle with a maximum fiber type will support a higher level of tension than an untrained muscle. A muscle that is trained to develop a high level of tension will have a lower level of tension when it is fatigued. The active tension-length curve is the sum of the passive and the active tension components.

The fundamental muscle mechanisms of the muscles in the body are determined by the length and the tension of the muscle fibers. The length of the muscle is determined by the degree of sarcomere overlap, which depends on the sarcomere length. The tension of the muscle is determined by the degree of sarcomere overlap, which depends on the sarcomere length. The tension of the muscle is determined by the degree of sarcomere overlap, which depends on the sarcomere length.
...continue reading...
anchor

The range of the launch configuration. By contrast, the design and structure of the rocket are determined by the exhaust pressure and the mass flow rate. The efficiency of the rocket's design is a function of the rocket's mass flow rate and the mass flow rate during the boost phase. The exhaust pressure is determined by the mass flow rate during the boost phase.

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The effective exhaust pressure is a function of the mass flow rate during the boost phase. The effective exhaust pressure is a function of the mass flow rate during the boost phase.
MUSCLE ACTING WHILE LENGTHENING

The model is used to calculate the two-tissue parameters described in the introduction of the model may be calculated in order to fit the experimental data. The model parameters are determined through an optimization process. The optimization process is carried out using a genetic algorithm. The optimization process is performed on a computer, and the results are used to determine the values of the parameters. The model is then used to predict the behavior of the muscle under a variety of conditions. It is found that the model accurately predicts the behavior of the muscle under a variety of conditions, and that the model is able to fit the experimental data with a high degree of accuracy.

Time Course of Active State in a Twitch

In order to achieve a steady state, the muscle must be allowed to relax completely. This is achieved by applying a constant length to the muscle, which is then allowed to relax. The muscle is then allowed to relax again, and this process is repeated until a steady state is achieved. The muscle is then allowed to relax, and the process is repeated until a steady state is achieved.

Fundamental Muscle Mechanics
Difficulties with the Active State Concept

Substituting for the pressure one gets

where the ?? was not defined independently when nonlinear parameters were

and the ?? were all constants. The shape of ??, ??, ??, ??, ??, and ?? were all constants, The shape of ??(??) and ??(??) considered the active state in a step function substitution method based on the response of a relaxed muscle to a step single measurement of the moment of inertia. The moment of inertia from a known fit is possible to calculate the time course of the active period from a

requiring to the model of ??(??) and ??(??) the parameters ??, ??, ??, ??, ??, and ?? are.

only ?? is to increase from ?? to ?? of the moment.

From the time of stimulation the active state lasts for up to 12

reasons discussed above. Experiments found that after a latency period of about 12

hours the tension developed. The active state became prominent. For the relative time delay at that moment for the

position of the tension as well as at the peak of the peaks must be understood as

Fundamental Muscle Mechanics
Summary and Conclusions

In summary, this chapter has introduced the fundamental principles of mechanics, focusing on the equilibrium of particles and systems. The concepts of force, moments, and torques have been discussed, along with their mathematical representations. The chapter concluded with an overview of the applications of these principles in engineering and physics, emphasizing the importance of understanding these fundamentals for solving real-world problems. Future chapters will build upon these concepts, exploring more complex systems and scenarios.
Problem 2

\[ \frac{h_{\text{peak}} - h_{\text{second peak}}}{L} = \frac{1}{(\gamma_{xy} - 1) L} \]

Here:

\[ (\gamma_{xy} - 1) L = \frac{1}{y} \]

\[ y = L \]

Therefore, the force \( F \) seconds after \( t \) is

\[ (\gamma_{xy} - 1) L = (F + C) L = (t + C) L \]

\[ F = C \]

After the \( L \) goes off, the force decays according to

\[ (\gamma_{xy} - 1) L = \frac{1}{y} \]

Since the step stays on for \( t \) seconds, the force level when the step goes off is

\[ (\gamma_{xy} - 1) L = \frac{1}{y} \]

If \( L \) is limited on a step, the initial conditions are \( L = 0 \). The solution

\[ L = \frac{y}{H} + L \]

Substituting into the equation describing the force generator plus dashpot,

\[ \frac{dL}{dy} = \frac{y - L}{H} \]

Differential the last equation above and substituting it into the first,

1. \( L \) is constant: \( \frac{dL}{dy} = 0 \)
2. Force generator plus dashpot: \( \frac{dL}{dy} = y \)
3. System: \( \frac{dL}{dy} = 0 \)

Assume \( C = 0 \), then the time that the controllable mechanism

Plant the answer to part (a) schematically.

Fundamental Muscle Mechanics
2. Device an expression for the values $T$ and $\gamma$ which maximize the power produced.

3. Show by substituting definitions of $\gamma$, $T$, and $X$, that eq. (1.4) is the

**Problems**

1. **Solution**

\[ (1 \gamma - x \gamma)^{\gamma} - \gamma \gamma^{\gamma - 1} = I \]

\[ \gamma \gamma^{\gamma - 1} + x \gamma^{\gamma - 1} = I \]

Finally,

\[ \gamma \gamma^{\gamma - 1} + x \gamma^{\gamma - 1} = I \]

Hence,

\[ \gamma \gamma^{\gamma - 1} + x \gamma^{\gamma - 1} = I \]

In model (a),

\[ \gamma \gamma^{\gamma - 1} + x \gamma^{\gamma - 1} = I \]

\[ \gamma \gamma^{\gamma - 1} + x \gamma^{\gamma - 1} = I \]

Substituting in the first equation for $\gamma$,

\[ \gamma \gamma^{\gamma - 1} + x \gamma^{\gamma - 1} = I \]

Substituting the first equation for $\gamma$ and use the second equation

\[ \gamma \gamma^{\gamma - 1} + x \gamma^{\gamma - 1} = I \]

Solving the first equation for $\gamma$ and using the second equation

Fundamental Mecanics, 0
the active state duration in single muscle or a series of muscles? Explain.

5. In human's method be used to find the maximum of the time course of

\[
\int_{-\infty}^{\infty} f(t) dt
\]

Rest in that the initial tension \( f(0) \) after the stretch event

step of active state force. \( f(0) \) is the initial tension \( f(0) \) is given at the same time as a

problem when a quick stretch of amplitude \( A \) is given at the time in the model of the first solved

4. Derive an expression for the tension in the model of the time course of the active state:

\[
\text{tension} = \int \text{parameters of the model} \times \text{duration of the}
\]

\text{muscles. Use the model of solved problem to calculate the tension in the muscles.}

have shown that it is higher in fast muscles than in slow ones and it depletes in

Fundamental Muscle Mechanics
An identified model for human wrist movements

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Summary. We have performed tests to find the mechanical properties of the hand and muscles driving wrist flexion and extension, and have identified parameters of a model. The hand acts as a nearly pure inertial load over most of its range of motion. It can be approximated as a rigid body rotating about a single axis. Viscosity of the wrist joint is negligible. Passive elastic torques are also small, except at extreme wrist angles. We measured torque as a function of wrist angle for maximum voluntary contractions, and angular velocity as a function of load. The torque/velocity curves for shortening muscles are well approximated by a Hill equation. To measure the "series elasticity" of the muscle equivalents, we imposed step changes in torque. The series stiffness is a monotonically increasing function of the preload, or "active state", in the Hill sense. We discuss the relationship of the measured parameters to properties of isolated muscles. To see the implications of the model structure for the "inverse problem" of identifying motor control signals, we simulated four models of different complexities, and found best fits to movement data, assuming simple pulse-shaped inputs. Inferred inputs depend strongly on model complexity. Finally, we compared the best fit control signals to recorded electromyograms.

Key words: Parameter identification - Voluntary movement - Human wrist

Introduction

The wrist is a popular system for the study of single-jointed limb motor control. Hand position is finely controlled, and the hand and muscles are accessible for measurement and perturbation. Electromyographic (EMG) responses to imposed torques (Calancie and Bawa 1985a; Jaeger et al. 1982), reflex stiffness and viscosity (Gielen and Houk 1984), and EMG patterns in voluntary movement (Sanes and Jennings 1984; Calancie and Bawa 1985b, Litvintsev and Seropyan 1977) have all been reported.

Most papers have analyzed wrist responses without explicit models, though some investigators (Stiles 1983, Lakie et al. 1984; Stein et al. 1988) have characterized the wrist as a second order system. Explicit models are necessary if movements are to be simulated, or if control patterns are to be inferred from recorded movements. The recorded movement may include position and its derivatives, and net torques. The best available measure of a "control pattern" is often the electromyogram. The inference step is, in effect, an inversion of the operator that takes control pattern to movement (e.g. the computation of Hannaford et al. 1985). We hypothesize that the inferred control pattern should depend strongly on the operator being inverted (the model).

We have identified parameters of a fourth order lumped-parameter model (Fig. 1) for studying wrist movements. Our model includes, first, the mechanical properties of the hand, the wrist joint, and the passive muscles.

\[ \begin{align*}
\text{B(uf,xt)} & \quad \text{B(ue,xe)} \\
\text{kp} & \\
\text{kc(uf)} & \\
\text{m} & \\
\text{kt} & \\
\text{kc(ue)} & \\
\end{align*} \]

\[ x_f \]

\[ x_e \]

\[ t \]

Fig. 1. The simplest model capable of the behaviors we measured includes an inertial mass \( m \) and a passive elastic element \( kp \) pulled by two muscle equivalents. Each "muscle" is composed of a force generator \( u(t) \) in parallel with a velocity- and active state-dependent dashpot \( B(u, x) \). These two are connected in series with two springs. One, \( kce(uf) \) is active state-dependent, and the other, \( kt \), has constant stiffness.

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Torques due to muscular activity are lumped into effects of two antagonistic "muscle equivalents". This lumping of forces produced by many muscles is justified, in part, by the fact that the relationship between torque and velocity, torque and angle, and series stiffness to preload are those one would expect from a single muscle. There is also some EMG evidence that all the wrist flexors turn on and off at the same time during rapid wrist movements (Litvintsev and Seropyan 1977).

We have not attempted to identify some properties. Activation and relaxation kinetics are the subject of a separate investigation. The inputs to our model are therefore the forces that the contractile element would develop, if the force/velocity curve played no part—Hill's "active state". Nor have we attempted to characterize a torque/velocity relationship for activated lengthening muscles.

Having identified parameters, we tested the model by simulating a fast wrist flexion. An optimization algorithm found the four-pulse input (two pulses per muscle equivalent) that minimized the difference between the simulated and actual wrist positions. We compare the derived "active state" pattern to electromyograms. Simulations of three simplified models test the hypothesis that inferred control pattern depends strongly on the complexity of the model.

**Methods**

**Subjects**

Subjects were three healthy adults between 21 and 40 years old. All protocols were approved by the Committee for the Protection of Human Subjects at the University of California, Berkeley.

**Apparatus**

The static parameter identifications were performed using a load cell (Sensotec Model 34) attached to a handle at one end and fixed to a table top by a large bolt at the other. Angles were varied in the static experiments by moving the bolt to one of several holes on a circle whose center was the hand (Fig. 2a).

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*Fig. 2a, b. Apparatus used in the experiments. a Series stiffnesses, force/velocity characteristics, and voluntary movements were measured using a handle connected to a motor with position, torque and acceleration sensors. The computer-controlled motor produced constant torques and torque steps for the stiffness measurements. Constant torques for the torque/velocity determination were provided by hanging weights over a pulley at the edge of the table (not shown). Electromyograms (electrodes shown) were recorded in all experiments. b Passive and active torque as a function of angle were measured by with a force transducer connected to a handle placed at many positions around a circle centered at the hand.*
The other parameter identifications and movement experiments were performed using a permanent magnet electric motor (Electrocraft Model 720), digitally controlled. The motor is mounted on the bottom of a small table, with the top of its shaft nearly flush with the table top (Fig. 2b). A rigid aluminum handle is mounted on top of the armature. The handle consists of a fixed plate and an adjustable one, between which the subject’s hand is firmly held. The subject thus flexes and extends the open hand about the wrist axis, which is coincident with the motor axis. Wooden blocks on top of the table constrain the subject’s forearm.

A potentiometer, an accelerometer and a torque transducer were added to the motor, and used to measure the movements. The motor has low inertia, but can produce a large torque (up to 6 Nm) and has a rapid response (mechanical time constant 4 ms). Driven by a power amplifier, it produces torque proportional to input voltage. We used this feature to produce step changes in torque to identify the series elastic elements.

For identification of the force/velocity characteristic, hanging weights were attached to the handle by a wire (bicycle brake cable) leading over a pulley at the edge of the table.

Electromyograms

Surface electromyograms were recorded during all of the tests. The electrodes were pairs of silver/silver chloride disks 8 mm in diameter, applied to the skin over the flexor carpi radialis and the extensor carpi radialis (Delage and Perotto 1981), 2 cm apart. We checked for crosstalk by looking for activity in each channel during a maximum contraction of the opposite muscle group, and moved electrodes if significant crosstalk was found. Exogenous signals from finger flexors and extensors were eliminated by positioning the hand and handle, thus avoiding any possibility. Signals were differentially amplified 2000–5000 times (Grass P511f), and bandpass filtered (half amplitudes at 10 Hz and 0.3 kHz). Sampling rate was 500 samples per second.

Results

Passive hand and wrist act as an inertia

The center of rotation of the wrist was found geometrically. The subject grasped a pencil and rotated the wrist through the full range of motion, with the forearm fixed on the table. Construction of the center by perpendicular bisectors of chords of this arc showed that the hand rotates about a single fixed center over most of the range of motion.

We measured the inertia of the apparatus by computing the acceleration imparted by a constant torque (0.375 Nm) applied by the motor. Digital differentiation of the position yielded a velocity, to which we fit a straight line by eye. The slope of this line gave acceleration. The inertia of the handle plus armature was \(7.12 \times 10^{-3}\) kgm\(^2\). We also added known masses (500 g and 1 kg) to the handle at a fixed radius (0.1 m). Comparing the velocity for these known added inertias to that with the relaxed hand in the apparatus (Fig. 3), we found that the inertia of the 0.4 kg hand was \(3.9 \times 10^{-3}\) kgm\(^2\). The measured inertia was close to that estimated from the hand mass, approximating the hand as a rigid parallelepiped, accelerated about one end (see discussion). Hand volumes were measured by water displacement, and masses estimated assuming a constant tissue density of 1 g/cm\(^3\). Subject’s hands varied in mass from 0.3 to 0.4 kg.

![Fig. 3. Velocity vs. time for the motor connected to the handle and various loads (from top): handle alone, handle with relaxed hand, handle with 500 g mass 10 cm from axis, and handle with 1000 g mass 10 cm from axis. Up to about 0.1 s, the hand acts as a simple inertia.](image)

Velocity was linear (Fig. 3) up to approximately 100 ms (the latency of the myotatic reflex EMG is about 40–50 ms, and EMG to force production takes an additional 35 ms), so the passive hand could be viewed as a purely inertial load at the velocities measured.

Passive elastic torques are small over ±40 degrees

We measured the passive forces (produced by the relaxed hand) at intervals of 0.3 radian (17.5 degrees), over the full range of movement. The subject placed the knuckles against a handle attached through a load cell to a bolt, which was fixed to the table. The direction of pull was varied by moving the bolt to holes in the table top, equally spaced on a circle centered at the hand.

Passive torques were less than 0.1 Nm over the central 80 degrees of each subject’s range of motion (Fig. 4). At either extreme of the range, torque rose rapidly.

Maximum torque is nearly independent of wrist angle over the same range

The maximum voluntary torque exerted by a subject was also nearly independent of angle over the central 80 degrees of the range of motion (Fig. 5), but declined sharply at each end of the range (to 20–25% of the maximum).

Torque-angular velocity relationships are well fit by Hill curves

We measured maximum velocity at a variety of fixed loads. The fixed torque was set by hanging a weight over the table edge, the weight attached to a cable which passed over a pulley to reach its attachment at the handle (ala Wilkie 1950). Subjects were instructed to move the
weight as fast as possible, and were told the peak velocity after each movement. Random ordering of the loads and ample rest periods between pulls reduced the likelihood of fatigue.

Wilkie (1950) noted that such a hanging weight apparatus produces both the constant torque due to the weight and a time-varying acceleration torque. His apparatus included a special lever system to minimize movement of the weight, and therefore the inertial torque. At the lightest loads, his subjects could not reach a maximum velocity (zero acceleration) within the range of motion. He therefore calculated (his appendix A) the theoretical velocity the arm would have reached, if it had no inertia, and scaled his velocity measurements accordingly.

Our subjects were able to reach and hold a maximum velocity before the end of the movement, even with the lightest loads. Perhaps the difference between wrist and forearm is due to the smaller inertia of the wrist, relative to the maximum torques produced by the muscles (Lehman and Stark 1983). In the worst case, the fastest movements, velocities peaked and stayed constant (to within 3%) for a period of 50 milliseconds. An accelerometer built into the apparatus verified that acceleration was indeed small during a time interval surrounding the time of peak velocity. During this peak, there was, of course, no acceleration and no inertial torque.

In our experiments, as in Wilkie's, the best argument for constant excitation during the movement is the consistent peak velocity we found at each load (Fig. 6). "The constancy of the mechanical response at all times makes it very likely that the excitation does remain constant, but a direct investigation has also been made by recording electromyograms during the movements" (Wilkie p. 272). Our electromyograms showed consistently high levels in the agonist around the time of maximum velocity, and minimal (relaxed) levels of electrical activity in the antagonist.
Fig. 6: Maximum voluntary angular velocity as a function of force for three subjects (bottom row, subject 1, 2, 3). Solid lines represent best-fit Hill hyperbolae. Summary curves (bottom row) compare Hill fits for all subjects.
Isometric maximum torques varied between subjects, but were consistently about twice as large for flexors as for extensors (Fig. 5, 6). Given feedback, subjects were able to produce consistent maximum velocities at each load (Fig. 6).

The Hill force-velocity relationship:

\[(P + a) (V + b) = (P_0 + a) b\]

was fit to each subject's data, varying \(P_0, V_{\text{max}}\) and \(a\) as free parameters. Least squares fits for each subject had errors of 1.1 to 5.9%, with \(a/P_0\) between 0.28 and 0.41. \(V_{\text{max}}\) varied less than did \(P_0\) between subjects, ranging from 15.3 to 26.2 radians per second.

Series elasticity – preloaded wrist responds as a mass-spring

We measured the series elasticity of the muscle equivalents by a quick-release protocol (Jewell and Wilkie 1958). The subject first held a fixed position by producing a constant torque against the motor. The constant preload was intended to produce a constant “active state”. The motor’s torque was then stepped to a value 0.75 Nm lower than the preload. The experiment was repeated 6 times at each preload, at preload varying from 0.75 to 6.0 Nm.

The response of an isolated muscle to a quick change in load is well known: a rapid change in length attributed to the series elastic element, followed by a slower component attributed to the interaction of the contractile element and the series elasticity. If the hand responded as a mass attached to a spring, the wrist angle would be

\[\chi(t) = \frac{F}{k} (1 - \cos \omega t)\]

where \(F\) is the constant torque after the step, \(k\) the spring constant, \(\omega = \sqrt{k/m}\). The wrist angle was well approximated by a cosine over the first 40 ms (Fig. 7). We measured both the frequency of this cosine and the max excursion, and deduced the short-range stiffness. Plot stiffness as a function of initial torque (Fig. 8) we found that the short-range stiffness was proportional to \(t_0\) for small initial torques. The slope was about the same (0.9 to 1.8 rad\(^{-1}\)) for each subject and for each muscle.

Measurement errors became significant at the larger loads (smaller deflections \(F/k\)).

**Discussion**

**Passive properties**

Over the middle of the range of motion (40 degrees c direction from the resting position) the hand acts nearly pure inertial load, moving about a single axis.

The inertia could well be estimated, approximating the hand as a rectangular parallelepiped of mass \(M\) length \(a\), rotated about one end. The parallel axis theorem then gives the inertia:

\[I = Ma^2/12 + Ma(a/2)^2 = Ma^2/3\]

For the subject whose hand was measured, the mass 0.4 kg, the measured \(I\) was \(3.9 \times 10^{-3}\) kgm\(^2\), so \(a = 1\). The length of this subject’s hand was 18 cm.

That viscous and passive elastic torques are shown in the constant torque input experiment (Fig.

the cosine fit to the response to quick release (Fig. 7)

the direct passive stiffness measurement (Fig. 8).

The resonance data of Lakie, Walsh and W. (1984) yields indirect, but fine estimates of these
so the passive elastic torque is four orders of magnitude smaller. Elastic torques are therefore comparable to active state only during posture. In a very rapid movement, wrist velocity might reach 10 radians per second, so viscous torques would peak at around $10^{-1}$ Nm, or 1/100 of $P_0$. Joint viscosity is therefore never significant.

Active properties

The shape of the active torque vs. angle curve depends on the subject (Fig. 5), but maximum torque generally shows only a weak dependence on angle.

The equivalent flexor and extensor have torque/velocity curves like force-velocity characteristics of isolated muscles. This, and evidence that the muscles are turned on and off at the same time during a variety of movements (Litvintsev and Seropyan 1977), justify lumping the flexors and extendors into two single muscle equivalents.

Series elasticity

Our imposed step reductions in torque produced trajectories well-fit by cosine curves (Fig. 7), as would be the case if the inertia of the hand were attached to a simple spring.

The cosine fit is good over a surprisingly large angle, given that the short-range stiffness exists only over a few percent of the rest length of the muscle. We therefore estimated the relationship between angular change and muscle length change from the geometry of the wrist joint (Fig. 9a). Given the angle of wrist rotation $\theta$, the radius from the center of rotation to the tendon insertion $r$, and the distance from the center of rotation to the carpal tunnel $w$, one can compute the length $L$ of tendon from the carpal tunnels to the tendon insertion on the palm by the law of cosines. If the wrist angle changes by $\Delta \theta$, one can compute the corresponding change in length $\Delta L$ by a second application of the same law. Assuming that $\Delta \theta$ is small, the change in length turns out to be

$$\Delta L = w r \Delta \theta \sin \theta / L$$

From a dissection, we estimated $w$ to be about 2 cm, and $r = 7$ cm. For small movements about resting position, $\theta$ is about 90 degrees, $\sin \theta \approx 1$, and $L \approx r$. Then

$$\Delta L \approx w r \Delta \theta / L^2 \approx w \Delta \theta / r,$$

so a 1% change in tendon length would correspond to an angular change of about 0.01 radians, or about 0.6 degrees.

Cosine curves fit the position trajectories over a range of about 6 degrees, or a few percent change in muscle length (like isolated muscles, cf. Jewell and Wilkie 1958).

It is also possible to estimate the muscle forces from the torques measured at the hand using the geometry (Fig. 9b). Let $F$ denote the force exerted by the muscle and $f$ the projection of that force normal to the palm...
Isolated muscles have series elasticities composed of a tendon spring in series with an activation-dependent "cross bridge" spring. As the force rises, cross bridge elasticity becomes comparable to tendon elasticity, and series stiffness saturates (Morgan 1977). Our length measurements at high preloads were too noisy to allow us to determine whether stiffness saturated or not.

A rough estimate of tendon stiffness is possible. Young's modulus for tendon rises over the first few percent of strain, then levels off at about $10^9$ N/m² (Wainwright et al. 1982, p. 89). To convert the Young's modulus of the tendon to stiffness requires estimates of the cross-sectional area and the length of the tendon. From dissection, the diameter of the flexor carpi radialis tendon is about 4 mm, and it is about 20 cm long. Its stiffness is then about $8 \times 10^4 \pm 10^3$ N/m.

Another way to estimate the tendon cross section is to use the fact that the maximum stress exerted on a tendon is between 1 and $2 \times 10^7$ N/m². If the muscle exerts 1000 N, it should then attach to a tendon with cross-sectional area between 0.5 and $1 \times 10^{-4}$ m², i.e. a diameter of 4–8 mm. Such a tendon would have a stiffness of 0.8–1.6 $10^4$ N/m.

The largest stiffness we measured (at a preload of 6 Nm, or about 30% of $P_o$) was about half the estimated tendon stiffness.

**Model simulation**

To find the effects of the identified mechanical properties on a whole movement, and to see the influence of model complexity on inferred "active state" input, we simulated four models. The most complicated model was that shown in Fig. 1, where $M$ is the inertia of the hand and handle, $k_e$ (u) and $k_t$ are spring constants representing the crossbridge and tendon elasticities, $B$ (u, x) is a variable dashpot to account for the force-velocity relationship and $u'$ and $w'$ are the "active state" of each muscle, in Hill's (Hill 1938) sense. The identified torque/angle relationships were not included in the simulations, because their effects are small over the range of interest.

The partition of the series elasticity into tendon ($k_t$) and cross-bridge ($k_e$) components followed from the identification of the series elasticity from torque steps (Fig. 8). The tendon-like $k_t$ has the high constant stiffness we found at high active state. The elasticity $k_e$ was made proportional to active state in the model. The two elements were then combined in the model as series springs (that is, with compliances adding).

The torque/velocity characteristics were measured only for maximum "active state". In the models, we assumed that the given fractional active state scaled the force available from the muscle at each velocity in the same way. That is, the torque/velocity curve for half maximal active state has the same $V_{max}$ half the $P_o$ and half the torque at each velocity.

The approximations to the full identified model, in decreasing order of complexity, were: a fourth order Hill model in which the entire series elasticity was assumed to reside in the tendon, a linear fourth order model in
Table 1. Parameters used in simulations

<table>
<thead>
<tr>
<th>Param.</th>
<th>value</th>
<th>Units</th>
<th>Remarks</th>
<th>Used in model</th>
</tr>
</thead>
<tbody>
<tr>
<td>( m )</td>
<td>( 3.9 \times 10^{-3} )</td>
<td>Kg m(^2)</td>
<td>Inertia of hand</td>
<td>I, II, III, IV</td>
</tr>
<tr>
<td>( M )</td>
<td>( 7.12 \times 10^{-3} )</td>
<td>Kg m(^2)</td>
<td>Inertia of apparatus</td>
<td>I, II, III, IV</td>
</tr>
<tr>
<td>( P_{ot} )</td>
<td>21.0</td>
<td>Nm</td>
<td>Max torque, flexors</td>
<td>I, II, III, IV</td>
</tr>
<tr>
<td>( P_{os} )</td>
<td>12.5</td>
<td>Nm</td>
<td>Max torque, extensors</td>
<td>I, II, III, IV</td>
</tr>
<tr>
<td>( D_{b} )</td>
<td>0.035</td>
<td>s</td>
<td>EMG to movement delay</td>
<td>I, II, III, IV</td>
</tr>
<tr>
<td>( V_{\text{max}} )</td>
<td>20.0</td>
<td>rad/s</td>
<td>Max shortening velocity</td>
<td>II, III, IV</td>
</tr>
<tr>
<td>( k_t )</td>
<td>10.0</td>
<td>Nm/rad</td>
<td>Tendon stiffness</td>
<td>II, III, IV</td>
</tr>
<tr>
<td>( a/t_{o} )</td>
<td>0.25</td>
<td></td>
<td>Hill a</td>
<td>III, IV</td>
</tr>
<tr>
<td>( k_{cc} )</td>
<td>1.5</td>
<td>l/rad</td>
<td>CE stiffness per unit a.u.</td>
<td>IV</td>
</tr>
</tbody>
</table>

which the viscosities B were assumed constant, and a second order model consisting only of the inertia. Parameters for all the models are listed in Table 1.

We simulated each model in C on an IBM PC/AT. In each case, we allowed square-wave inputs, with two pulses each of agonist and antagonist "active state". The height of each pulse, its width, and the delay before the pulse were free parameters (12 in all). We used an optimization algorithm to minimize root mean square error between model wrist position and a typical wrist movement trajectory, varying the input pattern.

The prototype wrist trajectory is a voluntary flexion of 0.23 radians (13 degrees), made as fast as possible, and measured using the motor apparatus already described (Fig. 2b). One of our three identified subjects placed his hand in the handle and viewed a target consisting of two lines on an oscilloscope (Fig. 2). By flexing and extending the wrist, he moved a cursor across the screen, into the target area (10% of movement amplitude wide). After a random delay of 0–4 s, the target jumped 13 degrees. The subject was instructed to move as rapidly as possible to the target, as soon as it jumped. After practicing for 20 to 30 trials, all subjects were able to make consistent movements (maximum standard deviation less than 10% of the movement amplitude), with consistent EMG bursts. The EMG's in Fig. 10 are rectified records from the single movement shown.

**Simulation results**

Good fits for all the models required the active state be delayed from EMG onset. In each case, the delay was around 35 ms. This delay includes the time from de-

![Fig. 10. a Linear models compared to a wrist movement. From the top, wrist angle best-fit agonist pulses, rectified flexor EMG (inverted for visual comparison), best-fit antagonist pulses, and rectified extensor EMG for a single fast wrist flexion. Long dashes denote a second order model (inertia alone). Short dashes correspond to fourth order linear model & Hill model (long dashes) and full identified model (short dashes) compared to a wrist movement. Same axes as in a for the model with Hill force-velocity curves and constant series elasticities, and the full model, with partitioned series elasticity](image-url)
polarization of the sarcolemma to force production, so part is true delay and part is the complicated kinetics of excitation-contraction coupling.

For each model, it was possible to tune agonist and antagonist active state pulses so that the model position matched actual position fairly well (rms errors < 10% full range, see Fig. 10). However, the deduced active states were very different from each other.

For the model consisting of the inertia alone (Fig. 10a, long dashes), driven by the difference between agonist and antagonist torque, the best-fit input approximated the acceleration (the hand is well-approximated by an inertia).

The linear fourth order model (Fig. 10a, short dashes), has constant series stiffness and muscle viscosity. The series springs and mass constitute a resonant system whose ringing, damped by the dashpot, can be driven to fit the wrist position. Late in the movement especially, the resonance of this system, and not the “active state” forcing, account for the shape of the trajectory. Note the vanishingly short (15 ms) second agonist pulse, and the absence of late antagonist pulse.

For the Hill model (Fig. 10b, long dashes), “viscosities” depended on shortening velocity and on “active state”. For the model inversion, the dependence on active state is especially important. At low active state, the viscosities are small. The series springs and mass are therefore uncoupled from the force generator (Fig. 1), so the trajectory depends more on the forcing, and less on the mechanical resonance. For example (Fig. 10b), uncoupling means the second agonist burst must be longer to drive position to the final value after the undershoot. A sustained antagonist pulse overlapping the two agonist pulses does allow the springs to fit part of the trajectory.

When the series elasticity is in part active state dependent, the inertia is further uncoupled from the muscles at low force levels. Now (Fig. 10b, short dashes), the late trajectory is almost entirely governed by the forcing functions. The best-fit active state has two agonist pulses, now separated by a longer delay than for the other models, and a sizable second antagonist pulse. For this model, the widths of the four pulses are approximately equal, as are the pulse widths of EMG.

Thus the inferred “active state” depends strongly on the assumed model, and the measured EMG and the forcing functions correspond more closely for the fullest model. The activation-dependent elasticities we measured show large effects on the inferred forcing function late in the movement, when activation is small.

References

Litvinov, Seropyan (1977) Muscular control of movements with one degree of freedom. Avtomatika i Telemehanika 5:88-102