Calculation of Human Arm Stiffness using a Biomechanical Model

Master Thesis
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Abstract

A novel hand-arm system (HASy), being developed in DLR as an anthropomorphic system, requires as an input parameter the realistic instantaneous joint stiffness of human upper limb in different postures and movements. Because muscles in our musculokeletal system are the active component (actuator), joint stiffness is driven by muscle stiffness. In this thesis, the theoretical fundamentals in extracting skeletal muscle stiffness from Delft Shoulder-Elbow Model (DSEM), a 3D finite element model of human upper limb, are presented in this thesis. The basics of musculoskeletal biomechanics and also skeletal muscle properties were studied in order to understand the arm stiffness, especially the skeletal muscle stiffness.

In order to complete the discussion of human arm stiffness, the three parameters contributing to arm stiffness, namely the end-point stiffness, the joint stiffness, and the muscle stiffness, were reviewed and their relationship with each other is discussed. Firstly, the mapping from muscle stiffness to joint stiffness will need to be done because HASy requires the joint stiffness value. Second, the most available arm stiffness research deals with end-point stiffness. Thus, knowing the corresponding end-point stiffness from the calculated muscle stiffness is good for future data validation.

Finally, the potential applications of this study are proposed. This study has implication in both robotics and biomechanics fields. In fields of robotics, specific application in HASy and also the possible application in general robotics are discussed briefly.

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Finally, I would like to dedicate all my studies, works, experiences and success to my mother and my sister, who have always loved me and been supportive in my life. Also, I dedicate my passion in learning and in relating my knowledge to improve the society for my beloved father in memory, whose love and guidance will always be remembered.

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Human ingenuity may make various inventions, but it will never devise any inventions more beautiful, nor more simple, nor more to the purpose than Nature does; because in her inventions nothing is wanting and nothing is superfluous.

Leonardo da Vinci, fifteenth century

Chapter 1

Introduction

In this thesis, the stiffness of human arm, specifically the corresponding muscle stiffness, has been studied focusing on application for the new DLR Hand-Arm System (HASy). In order to do so, a biomechanical model of the musculoskeletal system of the upper limb[1] is used. The biomechanical model used here is DSEM (Delft Shoulder Elbow Model), developed by Prof. Van der Helm from TU Delft.

In the following, ***human arm*** corresponds to *upper limb* or *upper extremity* consisting the shoulder and shoulder joints, upper arm, elbow joint, lower arm, wrist joint and hand. It is a musculoskeletal system consisting bones, skeletal muscles and joints.

1.1 Motivation

Setting adjustable stiffness is a design goal of DLR HASy [15]. This implies that the stiffness is a given parameter. Given several inputs (e.g. muscle activation level, environment constraints, movement goal, etc), HASy should be able to adjust the joint stiffness. Aiming for an anthropomorphic[2] system, the joint stiffness must be realistic just like the human joint stiffness when the same inputs are given. Since joint stiffness is resulted from the antagonistic action of muscles, it is crucial to understand and quantify the muscle stiffness. Section 1.2.1 presents background knowledge about HASy, which is based on [15].

[1] Limb: jointed part of the human body. Human arms are upper limb and the legs are lower limb

[2] Resembling human form and behavior

Measurement of the stiffness properties of both muscle and joint could be done by doing *in vivo* experiments. Unfortunately, this method is troublesome (may be invasive), time costly, and without high accuracy, at least with the current state of measurement technology. Using an available and good biomechanical-musculoskeletal model would be an effective and efficient option. Several complex musculoskeletal models are available and able to estimate muscle and joint forces from the given motion-goal (*inverse dynamics*) or neural input (*forward dynamics*). Brief overview of the biomechanical principles and also biomechanical modelling in kinesiology[3] can be found in sub chapter 1.2.2.

1.2 Background

1.2.1 DLR Hand-Arm System (HASy), an anthropomorphic system

There are different motivations for researchers in robotic field in developing human-mimicking robots. One of them is certainly the attractive adaptability and flexibility of our very own body that seem effortless. Another motivation is to understand better the human system, ask ”why” the nature does so, ”wonder” if there is any better way to build an anthropomorphic system without exactly copying the complexity of human body, and so on. DLR HASy is being developed to mimic the human upper limb in order to study our musculoskeletal system and to improve safety in robotics. Human-robot interaction demands a system that is safe for both the human and the robot. An anthropomorphic system can adapt its stiffness and respond to the environment. Interaction between human and an anthropomorphic-robot would be like human-human interaction. The realization of the biomimetic robots would certainly widen the horizon of robotic applications.

The design goal of the new DLR HASy is to closely copy the kinematic and dynamic properties of human arm rather than the intrinsic structure. It is an anthropomorphic system. This means that HASy will behave, move and have the form like human but its intrinsic structure or architecture does not necessarily resemble exactly the one of human.

[3]the study of human movement

The human musculoskeletal system is antagonistic. A skeletal muscle is a attached to the bones through tendon, which is responsible in transferring the force. It is important to note that muscle can only pull, not push. Thus, antagonism is necessary. Antagonism means that each joint has at least two muscles, one to flex and another to extend joint (Figure 1.1). The two muscles work as counter. Due to this antagonistic actuation, there is a precluded high stiffness during physical contact. Also, the musculoskeletal system has an ability to adjust stiffness as a function of different parameters. In result, our joints are non-linear and intrinsically flexible.

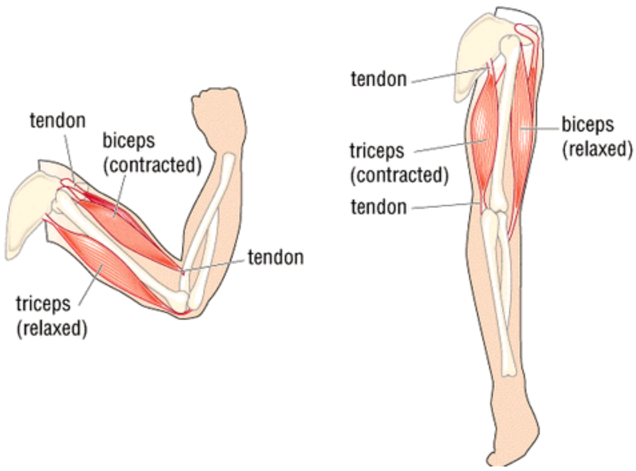
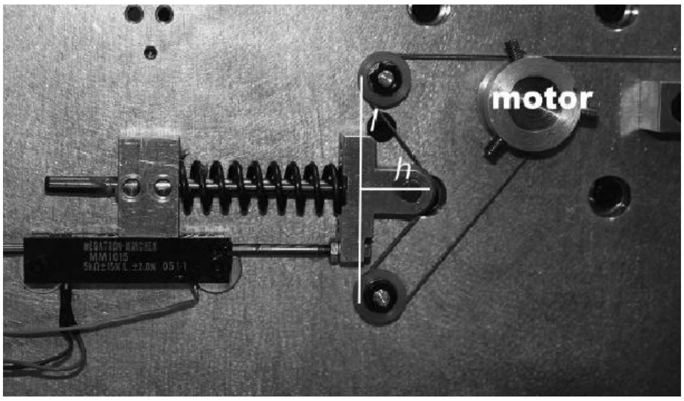
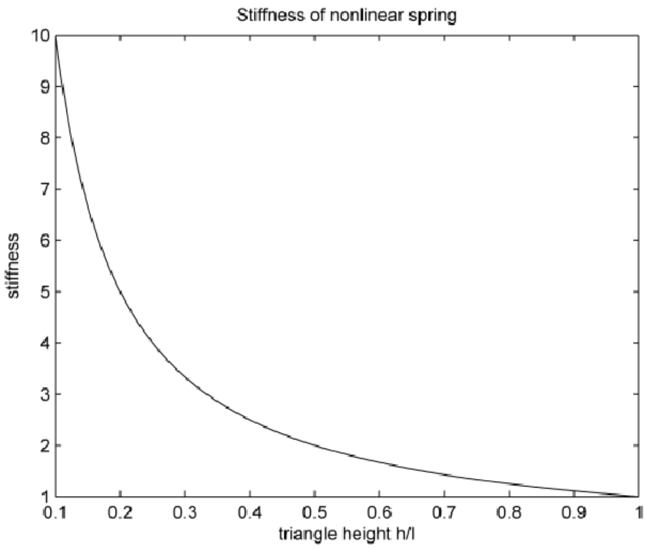


Figure 1.1: Bending and straightening of arm by antagonist muscles: Biceps and Triceps [1]

Likewise, HASy makes use of the antagonistic drive principles. Antagonism in HASy is realized by using two tendons to drive one joint (current prototype only has finger and wrist joints), one act as flexor and another as extensor. The joint would have variable and non-linear stiffness, just like the human joints. This is achieved by the variable stiffness actuating (VSA) system, which consists of several motors, each equipped with ***non-linear spring element*** to reproduce the realistic joint behavior. Each non-linear spring element consists of a linear spring which pushes the tendon (string), forming it into a triangle (figure 1.2a). The stiffness of the construction is determined by the ratio of the triangle height (h) and half base (l/2) as shown in figure 1.2b. The two motors, for example, can increase the total stiffness by pulling the antagonistic tendons in counter directions.



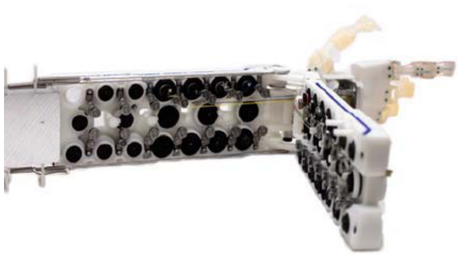
(a) The set up in testbed



(b) Stiffness as a function of h/l

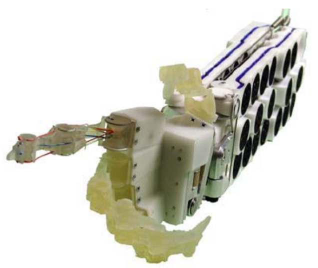
Figure 1.2: Non-linear spring element concept for HASy

In addition to the antagonistic actuating principles, HASy's important feature is to have weight, size and strength properties close to that of the human. This is achieved by applying the biomimetic components and also the use of a compact actuating system (figure 1.3). Here, the non-linear spring element is compact, not as shown in figure 1.2 that is only a test bed. Its biomimetic components are the endoskeleton and the bionic joints (Figure 1.3 c,d,e). Those who are interested in details of the finger joints developed for HASy are invited to read [15].



(a) compact motors

(b) tendons



(c) HASy's lower arm-hand Stereolithography prototype

(d) Index finger prototype

(e) Index finger in actuator testbed

Figure 1.3: HASy prototype, the lower arm-hand prototype has only the index finger as a moving part

1.2.2 Introduction to Biomechanics

Biomechanics, in general, is an interdisciplinary academic field that applies the mechanical principles in the living organism. It is a mixture of different basic knowledge, namely mathematics, physics, chemistry, mechanics (figure 1.4): statics, dynamics, and fluid mechanics, biology and medicine, neurophysiology, and behavior science. The more general scope of biomechanics deals with the whole range of living organism, from plant to animals. The more specific scope of biomechanics deals with human movement. The readers are expected to be familiar with the terminology of rigid body mechanics i.e. statics and dynamics (kinematics and kinetics).

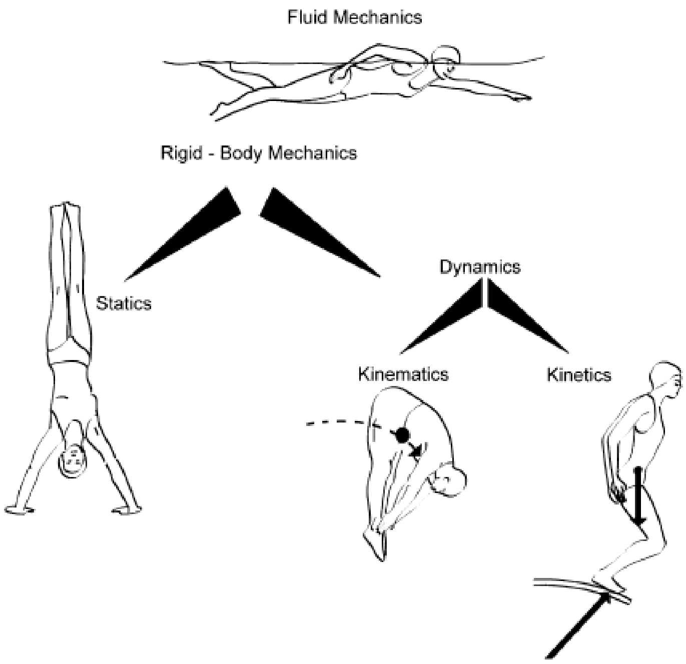


Figure 1.4: The divisions of Mechanics used in biomechanics of human movement [18]

Different literatures define the basic principles of biomechanics differently, some are more complex than some other. There are nine underlying principles in biomechanics, according to [18]: *force-motion principle, free body diagram, force-time principle, range of motion, balance, coordination continuum, segmental interaction, optimal projection,* and *spin*. The subsequent principles are discussed more because they are important in this study and the terminology are revisited in the discussion of musculoskeletal system in Chapter2.

Force-Motion principle. This principle is basically based on the *Newton's three laws*. It generally states that when a body creates or modifies movement, there is an imbalance of forces acting on it. It is also important to mention *inertia*, that is a property of a body to resist changes in its state of motion. *Mass* (m) is a linear measure of inertia and *moment of inertia* (I) is the angular measure. When simulating the kinetics of the musculoskeletal system, this principle is applied. Stiffness of a body is defined by the applied force and the resulted deformation. The force of a body itself is equal to the measure of inertia multiplied with body's acceleration ($\sum F = m \cdot a$ for linear motion and $\sum T = I \cdot \alpha$ for rotational motion)

Force-time principle. This principle states that not only the amount of force but also the amount of time in which the force is applied effect the motion. The underlying mathematical explanation of this principle is the *impulse-momentum* relationship. Without going into detail, it is known that history of muscle contraction i.e. previous force and velocity affect the instantaneous muscle action. It is also interesting in the existing of fast-twitch fibers and slow-twitch fibers in our skeletal muscle, which are function optimized. Fast-twitch fibers are those that can do fast movement but are not able to produce large force. The slow-twitch fibers can create large force but in long period of time. Thus, the design of the nature satisfies this principle.

Range of motion(ROM) is the overall motion in a movement. Depending on the purpose of movement, there may be minimization or maximization of ROM required. A baseball pitcher taking a longer stride is increasing the range of motion of the weight shift (figure 1.5). Articulation system (muscle-joint) has also specific ROM that depends on the architechture (chapter 2). Increasing or decreasing speed can be effectively achieved by increasing ROM. The effect of ROM in motion is related to Force-Time principle because larger ROM consumes longer time than in smaller ROM.

Balance is the ability of a person to control the body position around a base of support. [27] addresses this principle as *stability*. According to [27], higher stability in human body is obtained as the center of mass is lower, the base of support is larger, the base of support is closer to the center of mass, and the mass itself is greater. In postural stability, muscle and joint stiffness plays an important role. This will be discussed more in Chapter 3.

Coordination Continuum. This principle says the goal of the movement determines the optimal timing of muscle actions or segmental motions. There are strategies of muscle action depending on the movement goal. High force as a goal would most probably imply simultaneous action of different muscles and rotations of joints. Low force but high speed movement implies sequential muscle and joint action. Most motion or motor skills falls in between

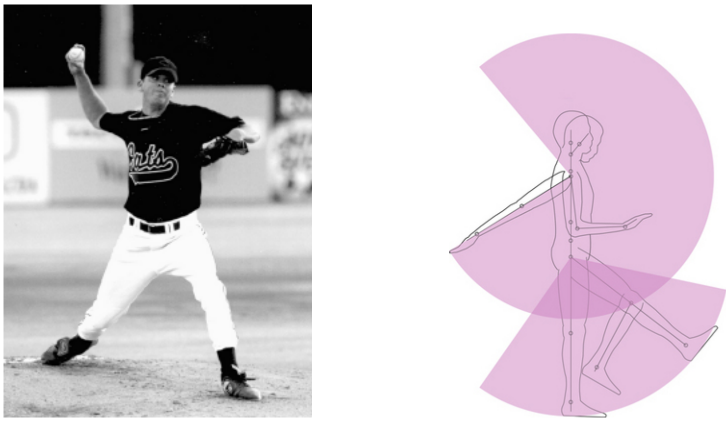


Figure 1.5: Baseball pitcher increasing its range of motion with the forward stride to increase body acceleration and eventually to accelerate the ball [18] and the side view of shoulder and hip range of motion [2]

simultaneous and sequential action. [27] interestingly states that production of maximum velocity requires the use of joints *in the order from the largest to smallest*, of which sport applications can be seen in hockey slapshot and in hitting golf ball. Thus, this principle says that there is management of the limb stiffness depending on the movement.

Segmental interaction states that forces acting on a multi-link system is transfered through the links and joints. Thus, there is transfer of stiffness parameters in human limb. The end-point stiffness is affected by the joint stiffness and also the chain architecture. Then, the joint stiffness is due to the stiffness of the attached muscles. Here, terminology to classify movements like open and closed chains in kinematic or kinetic analysis is introduced.

1.2.3 Preview of Limb Stiffness

Stiffness terminology in human limb is various. The three most relevant stiffness terms used in this thesis are *muscle stiffness*, *joint stiffness*, and *end-point stiffness*. Theoretically, joint stiffness is related to the muscle stiffness by the *muscle Jacobian matrix*[40] relating the muscle's length change and the joint's angular shift. Likewise, the end-point stiffness is related to the joint stiffness by the so called *chain Jacobian matrix*[40] relating the end-point's displacement with the joint's angular shift. These three stiffness values in our limb are discussed in this thesis (Chapter 3) to complete the whole discussion of limb stiffness. In this thesis, more focus is paid into the discussion regarding the muscle stiffness.

It is found that muscle stiffness is more intricate to be quantified. This is due to the fact that muscle stiffness has more complex physiological essence than the joint stiffness. For decades, people have realized that muscle has a spring like behavior and also somehow a viscoelasticity property. Specifically in this topic, skeletal muscle owns two components that contribute to its tension-length or force-displacement relationship. These two components are the active muscle fibers and the passive connective tissue (tendon and aponeurosis). In quantifying the muscle stiffness, it is important to discuss how the contribution from active and passive components is taken into account. Details on structure, functions, and properties of skeletal muscle are discussed in Chapter 2

Finally, the study of stiffness has implication in both biomechanics and robotics. Stiffness in our body does not only imply in the kinetics of the system, meaning to say how well the forces can be translated into displacements, but also in the stability. Stability is how well a system return to equilibrium point or trajectory after a perturbation is applied. Study of human postural and movement stability is very important in kinesiology for many applications e.g. sport medicine, orthotics[4] and prosthetics[5] study, robotics, etc. Specifically in robotics, the new trend of its development is toward anthropomorphic system. In the anthropomorphic automatic system, the system is not only intrinsically flexible but also non linear. Anthropomorphic robot is hoped to be desirable for the human-robot interactive systems. Also, anthropomorphic robotics will advance the manipulative ability. For space robotics, this would also mean higher safety and better performance.

1.3 Objectives and Scope

The two main objectives of this project are to study and calculate the muscle stiffness of the human upper limb using DSEM (Delft Shoulder-Elbow Model) as the biomechanical model. In order to be able to extract the useful information about stiffness (of both muscle and joint), literature review of the properties of muscle and also the relationship of muscle activation level

[4] A discipline that is concerned with design, development, fitting and manufacturing of *orthoses*, devices that support or correct musculoskeletal deformities and/or abnormalities of the human body.

[5] A discipline that develops *prosthesis*, an artificial extension that replaces a missing body part.

and the muscle stiffness were done. The study and application of relating the activation and also muscle stiffness is a subject of the neuromusculoskeletal study. In this thesis, more discussion and work were focused in the musculoskeletal system only. The study of muscle properties is a very broad topic; thus, the study was focused in the upper limb including its skeletal muscle system, i.e. the muculotendinous complex, and the muscle-joint system. In other word, biomechanics of the musculoskeletal system, especially the one of human upper limb, were reviewed. Also, in working with DSEM, the work was more focused in the program's subroutines and also output parameters that are more related to finding out the muscle stiffness value.

1.4 Thesis Outline

The Introduction chapter of this thesis discusses the general motivation of the project, the DLR HASy (Hand Arm System) and background knowledge of Biomechanics and also the limb stiffness topic. Chapter 2 discusses more in depth about human musculoskeletal system with highlights on the properties of skeletal muscle. The understanding and quantification of the limb stiffness parameters (the muscle, joint and end-point stiffness), as well as the relationship between muscle activation and muscle stiffness are presented in Chapter 3. Chapter 4 presents in detail the DSEM itself and the proposed application of human arm stiffness calculation. Finally, Chapter 5 summarizes the work and the future works to be done and discusses the application of this study in several fields.

Chapter 2

Biomechanics of Human Arm: Properties and Modelling of The Skeletal Muscle

The human arm is a musculoskeletal system. It consists of skeleton (bones), skeletal muscles and joints. Figure 2.1 shows the bones and the musculature of human arm.

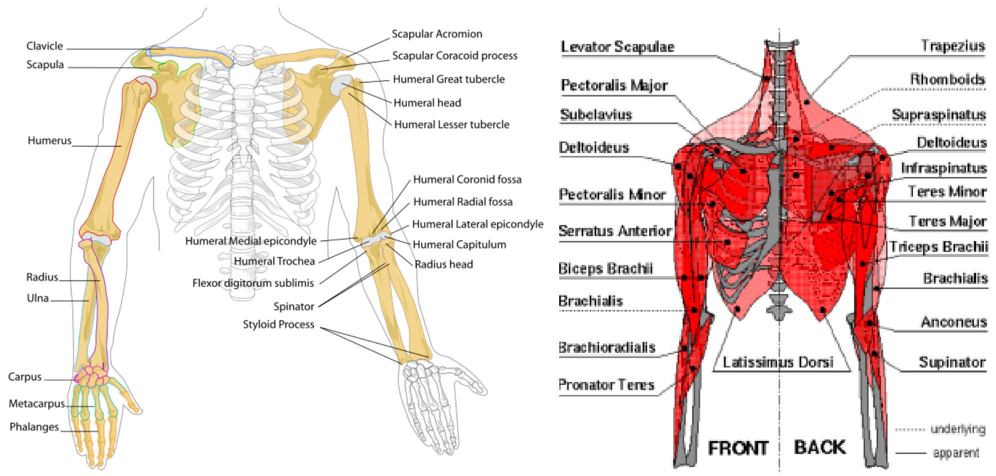


Figure 2.1: The bones structure and musculature of human arm

2.1 Structure and Functions of The Skeletal Muscle

There are three main types of muscle in human body, namely the skeletal muscle, smooth muscle, and cardiac muscle. When dealing with arm movement, only skeletal muscles are interesting. The skeletal muscle itself is unique because it is attached to bones, has multiple nuclei and peripherals, is striated[1], and can do both voluntary and involuntary contraction. Involuntary action is commonly called reflex action. A muscle contraction can be eccentric and concentric. An eccentric contraction is when a muscle lengthens while contracting. A concentric contraction is when a muscle shortens while contracting.

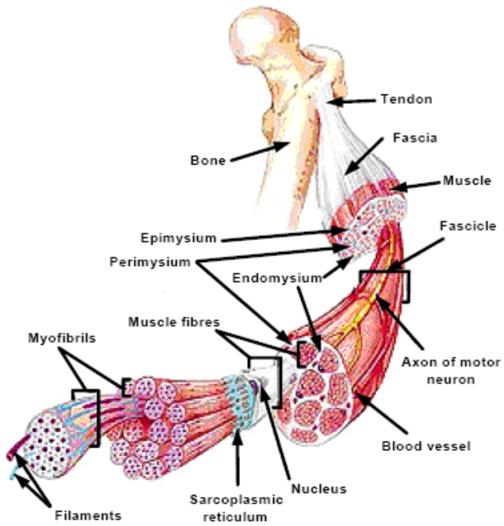


Figure 2.2: A skeletal muscle with many fibers surrounded by the connective tissue

The composition of a skeletal muscle is hierarchic (Figure 2.2). The muscle fibers (contractile elements) are surrounded by the connective tissues (collagen rich tissues). The voluntary contraction is actually happening in the shortening of sarcomeres, as mentioned in the famous "Sliding filament theory" proposed by Huxley in 1957 [20],[38], [11]. *Sarcomere* is a subunit of *Myofibril*. A Muscle fiber is formed by hundreds of Myofibrils. A muscle fiber is surrounded by Endomysium. Several of these sets form a *bundle* or *fascicle*. A group of bundles is surrounded by Perimysium. Epimysium covers the whole muscle and is continuous with tendon. Endomysium, Perimysium

[1] It consists of many fibers

and Epymisium are often called *aponeurosis*. Tendon, which is also a type of connective tissue, is placed at the two ends of muscle, attached to the bones, commonly called as *origin* or *proximal site* and *insertion* or *distal site*. In addition, there is an abundant network of nerves and blood vessels.

The skeletal muscle has significant functions in our body, especially in its role to create mobility. The functions of muscle's contractile element are **to move the body limb by creating motion** and **to provide strength by generating an active force**. The specific functions of connective tissues within a muscle are **to provide gross structure to muscle**, **to generate passive tension against stretch**, and **to transmit force to the bone and across the joint**. The skeletal muscle as a whole also functions **to protect joints by absorbing shock**.

Having mentioned the basic structure and functions of the skeletal muscle, a few important terms are introduced. An *Active force* is the force created by the *active* or *contractile element* in respond to the neural activation. The *passive tension* or force is created by the connective tissue. Thus, the connective tissue is often called as *passive element*. The term tension can be used instead of force because they are proportional (Tension = Force / Cross-sectional area).

In order to generate motions and/or stability, a chain of actions takes place in the musculoskeletal system. Skeletal muscles surround or wrap bones and cross the joints. Bones are connected via joints. When the skeletal muscle is activated by the innervating nerves, it would contract and thus, create force. Afterward,a moment is generated in the joint and the joint would be displaced (rotated, translated, etc). In other word, the motion is generated. This muscle contraction could also stabilize the joint, whether or not there is motion generated .

Specifically, the study of mechanical properties of the skeletal muscle is based on the interest in understanding the skeletal muscle function to create force, both actively and passively. Although skeletal muscles are responsible in contracting and generating force, without tendon the force would not be transfered onto bones and no joint rotation would be possible. Without the connective tissue i.e. tendon and aponeurosis (page 12 and 14), a skeletal muscle would not have the form and strength to do its functions.

2.2 Qualitative properties of skeletal muscle

Generally, muscles can generate force actively and also, using their *mechanical impedance*, provide passive force i.e. resistance to the imposed motion. It is known that they have spring-like properties and also velocity dependent properties. While the spring-like properties ensure stable mechanical behavior when interacting with the outer world, the velocity dependent forces help in damping undesirable joints oscillations [22]. Stiffness itself, as a known mechanical property, is related to the behavior of a material in resisting an applied force by undergoing deformation.

Here, the properties of the skeletal muscle are summarized. It is also imperative to distinguish the properties of the muscle fibers and of the connective tissue (i.e. tendon and aponeurosis).

- **Excitability**. A muscle fiber is capable to respond to a stimulus. There is a so-called *Activation and deactivation dynamics* that explains this property. Therefore, it is also called as the *active element*. Due to the excitability, the mechanical impedance is also a function of *activation level*. This gives the flexibility to our musculoskeletal system in modifying its mechanical characteristics based on the specific tasks or in changing (manipulating) the mechanical characteristics of the environment [22].

- **Contractility**. It is the ability of a muscle fiber to shorten with force and thus, is very often addressed as *contractile element*. The output of the activation and deactivation dynamic process becomes the input to the contractile element. The contractile element has specific *force-length* or *tension-length curve*. It is important to mention that in addition to force and tension, the term *torque* is also often used here. The joint torque is resulted by multiplying the acting muscle force with a certain *muscle moment arm*.

- **Extensibility and elasticity**. A muscle fiber can be stretched to its normal resting length and beyond to a limited degree. Muscle fiber is also able to recoil to original resting length after stretched. This property relates to the *active elasticity* or *active stiffness* of the muscle.

- **Viscoelasticity**. *Viscosity* or *dissipative property* is shown by the well-known *Force-Velocity* relation that is the dependency of the muscle force on the rate of change of muscle length and vice versa. The viscoelasticity is basically from the connective tissue. The connective tissue has also a force-length relationship that can be added to the one of the contractile element resulting in the total force-length relationship. This is the *passive elasticity*. For example, tensile test of tendon shows hysteresis during loading and unloading [22].

- **Non-contractility**. It is clear that tendon and aponeurosis are the passive components as they are not able to contract and create tension by themselves. Therefore, they are often called as the *passive elements*.

The force-length and force-velocity characteristics are also called as *functional properties* of skeletal muscle. This is due to the fact that the main function of skeletal muscle is to produce force. The functional properties show how well a muscle can create force, related to the muscle length and contraction velocity.

As a material, skeletal muscles are essentially incompressible, anisotropic, and, if a single fiber direction exists at each point, they may be considered transversely isotropic [6]. Due to the viscosity and activation dependence, skeletal muscles show non-linear properties. It can be concluded that a skeletal muscle has a complex behavior.

2.3 Muscle architecture

The mechanical properties of the skeletal muscle, which is our interest in studying the muscle stiffness, generally depend on the properties of the fibers and also the organization of the fibers or *muscle architecture*. For skeletal muscles, functional properties depend strongly on the architecture. The muscle architecture is explained by several parameters, namely *pennation angle, physiological cross sectional area (PCSA), muscle fiber length, tendon slack length* and *moment arm*. Before going in detail on formalization of muscle mechanical properties, the parameters defining muscle architecture are described here.

Line of action and pennation angle

To start with, a concept of *line of action* or *line of pull* should be introduced. A line of action is the line that joints the origin and insertion of the muscle and thus, the line of the acting muscle force. When there is an angle between the muscle fibers direction and the line of action, muscle is pennated with an angle called as the *pennation angle*. In our body, the pennation angle depends on functions and space. For example, a muscle for high speed but low force movements is unpennated. When a muscle is pennated, the number of fibers per unit volume is increased and overall contraction length is reduced. Therefore a pennated muscle is slower but produces higher force.

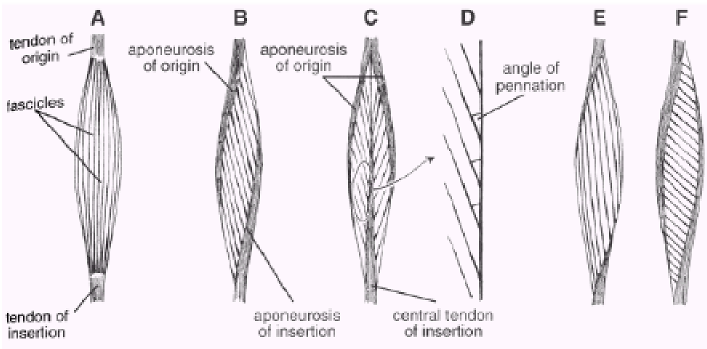


Figure 2.3: Muscle architecture defined by the pennation angle: A: Non-pennate, B,E,F: Unipennate, C: Bipennate with pennation angle shown in D.

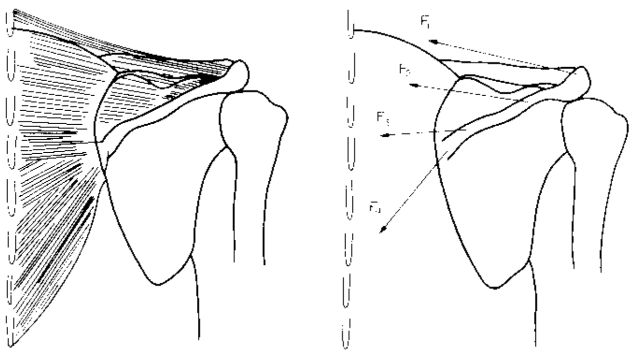


Figure 2.4: Muscle trapezius is represented with several force vectors (line of actions) to represent the laterally rotating torque [31].

In our musculoskeletal system, muscles are not only pennated but also with large attachment area, i.e. insertion and origin cannot be represented by a single point. Therefore, muscles with large attachment area are represented by several lines of action (Figure 2.4). A method has been developed by Van der Helm to define the force vectors so that the mechanical effect of the muscle is well represented [31]. In this method, the muscle is allowed to follow a bony contour. Also, form and size of the muscle attachments as well as the distribution of the fibers in the muscle are taken into consideration;

thus, it is appropriate to model the shoulder musculature accurately [31],[21]. Two other methods representing the line of action are the Jensen's *centroid line method* and Seireg's *straight line method* [35], [21]. The line of action representation options are available in figure 2.5.

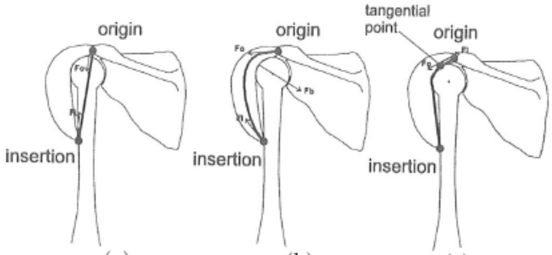


Figure 2.5: Muscle line of action from straight line approach (a), centroid line approach (b), and bony contour approach (c)

Moment arm

The concept of line of action brings another important concept that is the *moment arm*. It is the perpendicular distance from the line of action to the center of joint rotation (Figure 2.6). In mechanics, this term is often called as the *mechanical advantage* or *lever arm*. Thus, moment arm is an important parameter in defining the transfer from a muscle force into a specific joint torque.

The moment arm represents the *mechanical mapping* between the muscle and joint [38]. There are two bicausal mechanical mappings in muscle-joint system:

- between muscle force and joint moment
- between muscle velocity with joint velocity

It is understood that muscle velocity corresponds to muscle length while joint velocity corresponds to joint angle.

Mathematically, this mapping or transformation can be described by a Jacobian J_{mj} that is the differential mapping between muscle length and joint angle changes. Let n_m be the number of muscle degrees of freedom (DOF) and n_q for the joints, the following is applied [38].

$$\mathbf{d}x_m = J_{mj}\mathbf{d}x_q \qquad ; \qquad f_j = J_{mj}^T f_m \tag{2.1}$$

Where,

x_q : generalized joint coordinates (commonly angles)

x_m : the muscle lengths

f_m : muscle force associated with specific muscle DOF

f_j : generalized joint forces (usually moments) associated with each DOF

In other words, the transposed muscle Jacobian J_{mj} is the moment arm that is to be post-multiplied with muscle force in order to find the resulted moment or torque on the joint. Details on the Jacobian matrices of human arm as a kinematic chain are presented in sub chapter 3.2.

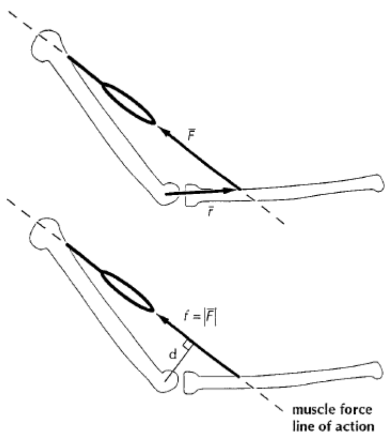


Figure 2.6: Simplified arm consisting one muscle, elbow joint, upper arm and lower arm. Force (**F**) vector is acting on the line of action. Considering planar movement, moment arm (**d**) can be multiplied by **F** to calculate the moment about joint rotation axis. The distance from center of rotation axis to the line of action of the muscle is represented by **r** (location vector). [25]

PCSA: Physiological Cross Sectional Area

PCSA is sum of all of the cross sections of the fibers in a muscle. For a non pennate muscle, PCSA is the area of the slice in the middle of the muscle, perpendicular to line of action (Figure 2.7 A). For a pennate muscle, the slice is taken perpendicular to the average of fibers direction (Figure 2.7 B,C). For a given volume, the pennate muscle would have larger PCSA than the non-pennate muscle.

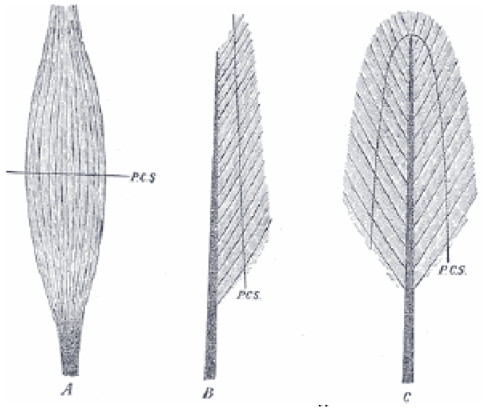


Figure 2.7: PCSA of non pennate muscle (A), unipennate muscle (B), bipennate muscle (C)

Fiber length and Tendon slack length

Skeletal muscle length is actually sum of the fiber length and the tendon length. Fiber length is proportional to muscle excursion and is related to the active contraction velocity (page 28 and 31). *Tendon slack length* is an important parameter in defining the muscle length and the muscle constitutive relationship, eventually the total muscle stiffness. By assuming a tendon to be totally stiff, the slack length can be neglected [14]. It is highlighted in [14] that the assumed value of tendon slack length determines the compliance and thus the force response of the actuator (muscle) as a whole. Thus, the accuracy in assuming the tendon slack length defines the accuracy of movement simulation models.

2.4 Mechanical properties of skeletal muscle

2.4.1 Quantification of properties via musculoskeletal system modelling

According to [5],[20], anatomical and biomechanical properties of skeletal muscle have been widely studied since the 1600s. However, quantification of these properties depends largely on experiments and mathematical or mechanical modelling of the skeletal muscle. In biomechanical modelling of the musculoskeletal system, the modelers usually aim to reproduce the properties of the musculoskeletal system with their models. Terminology in discussing the mechanical properties of muscle come from the early works in studying and modelling the musculoskeletal system.

The classification of muscle models is quite diversified. According to [11] the followings are four types of currently existing muscle models, which vary in complexity and also dimensionality:

- **Hill-type models** are the most popular muscle model known in biomechanics. Hill-type model was developed by A.V. Hill (1938,1950) based on experimental observations [11],[38]. This model aims to produce the mechanical properties of muscle. It has an active contractile element, a series passive-spring element, and a parallel passive-spring element to model the excitability-contractility of muscle fibers and the properties of passive-serial tendon and passive-parallel aponeurosis. Hill-type model is also called a phenomenological model [14], [38]

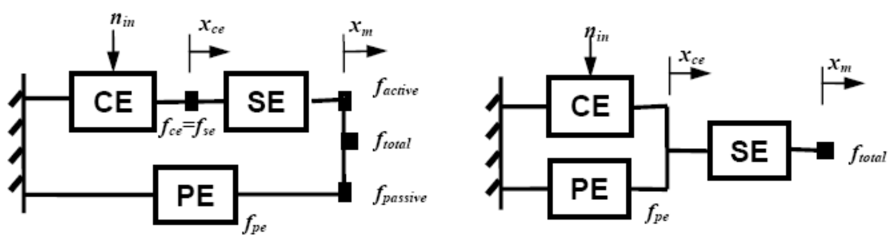


Figure 2.8: Two possible configuration of Hill's model showing the passive series element(SE), passive parallel element(PE), and slugish contractile element(CE)[38]. Neural input (n_{in}) is given to CE.

Most common form of the Hill muscle model is shown in 2.8left, with contractile element (CE) as the active contractile element being in series with series element (SE) and in parallel with parallel element (PE). SE and PE are passive elements that are represented by light damped springs [38]. The one on the right is an alternative form and has different constitutive relations. From the figure, it can be seen that the input to the model is the neural input (n_{in}) and the output is the total force (f_{tot}). From the model on the left, the total force exerted by the muscle is sum of the force from series element and the parallel element ($f_{tot} = f_{se} + f_{pe}$) with force of series element equal to the one exerted by the contractile element ($f_{se} = f_{ce}$).

It was first suggested by Hill in 1958 that when a neural input is maintained, a sudden change in force imposes also a sudden change of length i.e. spring-like behavior [38]. This leads to the derivation of spring constant simply as,

$$k = \Delta F/\Delta l \tag{2.2}$$

Taking the most common hill's model representation (Figure 2.8left), the Δl is just the x_m that is the summation of CE's and SE's displacement $x_m = x_{ce} + x_{se}$. When the neural input is changed, muscle show a damping behavior [38]. Also, linearization of the change of force and length can be done when looking at a very short time (sudden change).

- **Cross-bridge model** or Huxley-type model was first developed by Huxley and based on the cross-bridge dynamics [11]. This model could describe the force-velocity characteristics and the energy metabolism. Cross-bridge model could also be used to investigate the muscle force exerted in time. Huxley model is not as popular as Hill's model due to its complexity and great detail that are sometimes not worth when simulation of movement is the interest.

- **Morphological models** are based on the fact that muscle volume remains as it contracts and that the exerted force depends strongly in morphology of the muscle [11]. This model combines the geometrical properties of muscle and tendon i.e. length, pennation angle, etc. Morphological models usually predict the geometrical properties of muscles and the isometric force-length curve. An example is the modified Hill's model of Zajac that was published in 1989 and is used by the CHARM project [21]. Figure 2.9 shows the musculotendinous unit with the force-length and force-velocity curves.

  Pennation angle is the angle between the muscle fibers direction and the line of pull, which is through tendon in origin and insertion site. Further on pennation angle and other parameters defining muscle architecture is discussed in sub chapter 2.3.

- **Morpho-mechanical models** use finite element method (FEM) in order to take into account all the structural and mechanical properties of skeletal muscles [11]. It also assumes that muscle tissue is *continuum* in order to study the mechanism of force exertion.

Biomechanical model of skeletal muscle can also be classified into 1D, 2D, and 3D *continuum* muscle models, with 1D (one dimensional) model as the first developed and the simplest to the latest trend of highly complex 3D (three dimensional) models [6]. Hill-type and Huxley-type models are basically 1D, Morphological models have been developed in 2D and 3D, and

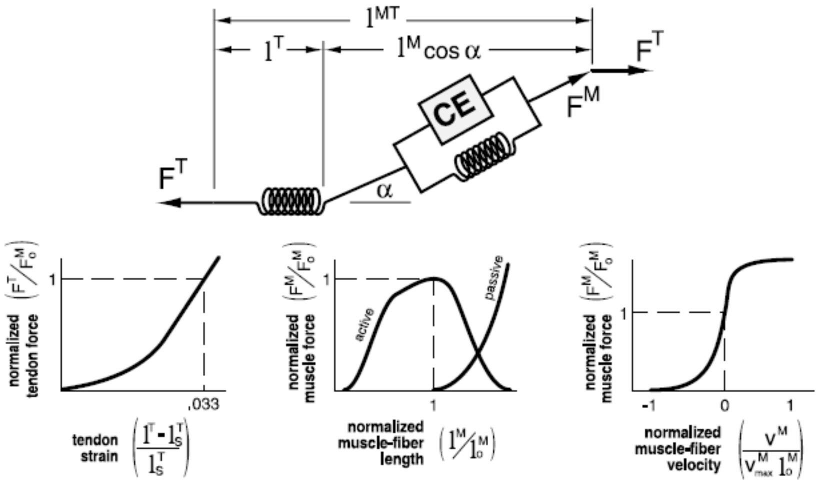


Figure 2.9: Zajac's musculotendinous unit model, published in 1989, taking the pennation angle α into consideration [9]

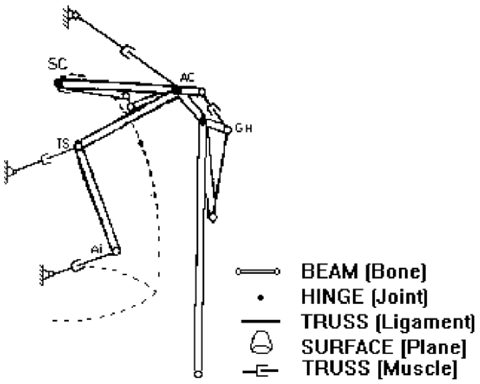
Morpho-mechanical models are commonly 3D. DSEM (Delft shoulder-elbow model) used in this project is a three dimensional morpho-mechanical models that relates muscle geometrical properties to force estimation in the shoulder-elbow system. Further details of DSEM is presented in Chapter 4.

It is worth mentioning that these models of skeletal muscle are based on ***continuum mechanics***. In continuum mechanics, the fact that a body consists of atoms that is discrete or non-continuous is neglected. In continuum mechanical models, *differential equations* can be employed in solving problems. There exist equations specific to the materials that are called the *constitutive equations*. The following sub chapter discusses the constitutive equations of skeletal muscle i.e. force-length and force-velocity relationship.

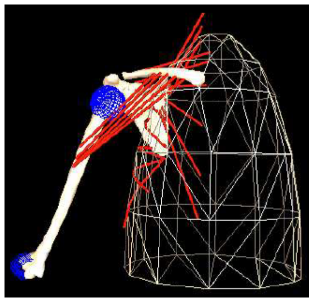
Morpho-mechanical modelling of the musculoskeletal system

Examining human body and movement is limited. One could not derive how and which muscles are working during movement just by looking at this movement. Obviously, current advance in bioimaging technology has supported the experiments positively. Another method is by modelling the system and experimenting different hypotheses using the hypothetical parameters, hoping to reproduce the behavior of the system like in real life.

Currently, there are several complex 3D (three dimensional) limb models available. Some of these models are especially human arm models, as shown in Figure 2.10, 2.11, 2.12. These models include bones, joints and muscles of human arm. An important part of musculoskeletal modelling is the modelling of skeletal muscle. Skeletal muscle modelling is a known subject in biomechanics and has been developed for decades. Skeletal muscle models ranging from the simplest to the most complex have been presented.

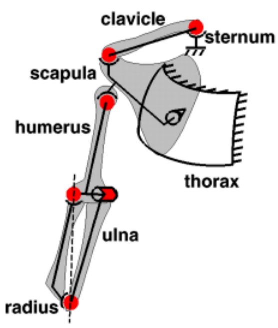


(a) Finite element representation

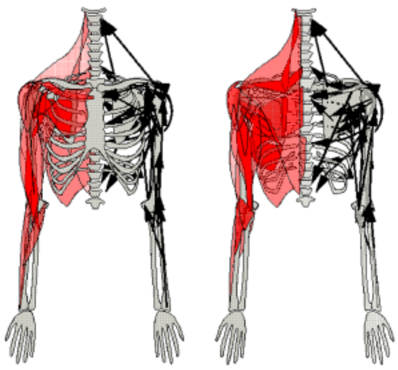


(b) Visualization with SIMM (Musculographics Inc.)

Figure 2.10: DSM (Delft Shoulder Model) before inclusion of elbow and forearm is an older version of DSEM (Delft Shoulder-Elbow Model)



(a) Skeleton model



(b) Muscle model

Figure 2.11: CHARM upper limb model [21]

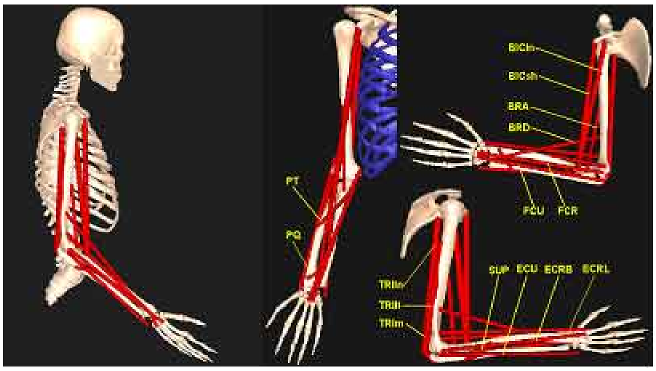


Figure 2.12: Musculokeletal modelling by Loeb and colleagues [3]

The popularity of musculoskeletal modelling is due to the fact that muscle and joint forces cannot be measured directly; thus, a model that is able to simulate motions and calculate the forces acting during the motions is required [29]. The specific goals of the development of the biomechanical model used in this work (DSEM) are presented in sub chapter 4.1. The followings are examples of the functions of biomechanical models, cited from [7], which demonstrate its importance:

- Analysis of clinical problems
  - diagnosis of disorders
  - improvement of current treatments
  - development of new treatments
- Insight into human function
  - muscle function
  - coordination
  - energy usage
  - muscle and joint forces
- Computer-assisted surgery
  - optimization of treatment for specific patient

It is clear that biomechanical modelling, especially of the human limb or musculoskeletal system, has mainly medical application. This project is eventually relating biomechanical modelling with the development of an anthropomorphic system.

Commonly, these large musculoskeletal models are able or aiming to simulate both inverse dynamics and forward dynamics. The inverse dynamics has recorded motions as input and is able to calculate the muscle forces and/or muscle activations by evaluating the motion equations. Motion equations in inverse dynamics is algebraic since position, velocity, and acceleration of the body is known. The forward dynamics takes in the neural input to the muscle and then integrating the differential motion equations to get the right motion.

As a musculoskeletal model reproduces either complete or specific properties of the musculoskeletal system, an anthropomorphic system can utilize the model to study and quantify these properties. If a musculoskeletal model can simulate movement and estimate the right muscle forces and eventually joint torques, it could be used to process and calculate the control parameters for an anthropomorphic system i.e. joint torques and rotations (joint stiffness), orientation, time step, and even sensory-feedback parameters. In addition, as the biomechanical models of musculoskeletal system are commonly computer programs, it should be more compatible to robotic systems.

2.4.2 Constitutive equations

The description of muscle biomechanical (or simply mechanical) properties have been studied extensively by experiments in model systems, where muscle sample is easily available i.e. animals, and in controlled environment i.e. with constant length or constant velocity [20]. According to [20], the mechanical properties of skeletal muscle are shown by the *Isometric Active Tension-Length* relationship, *Passive Tension-Length* relationship and *Isotonic Active Force-Velocity* relationship. This terminology is adapted in discussing the constitutive equations. Personally, the constitutive equations try to describe simply how elastic and how viscous the skeletal muscle is.

Constitutive relationship, mentioned as the force-length and force-velocity relationship, attempt to describe observed behavior of skeletal muscle by idealized lumped elements [38]. These lumped parameters have been introduced earlier, namely the *contractile* or *active element* representing the contractile and excitable muscle fibers, *series passive element* representing the passive connective tissue located in series with muscle fibers i.e. tendon, and *parallel passive element* representing the behavior of connective tissue that surrounds the muscle fibers i.e. aponeurosis. The first and most famous model of skeletal muscle, which introduces these lumped elements and the constitutive relationship, is the Hill's model.

Constitutive Relationship: Isometric Active Tension-Length and Passive Tension-Length

Isometric means constant length. Isometric active contraction is a *voluntary activated contraction* at which muscle length is maintained against a specific load (Figure 2.13). The constitutive relationship is obtained in experiment by allowing the muscle contract isometrically and recording the *peak isometric tension* of the specific length (Figure 2.14). The experiment is done in all range of muscle length, including above and below muscle rest length. Plotting the peak isometric tension against the length, one could see a dome shape of curve with optimum force at resting length (Figure 2.14).

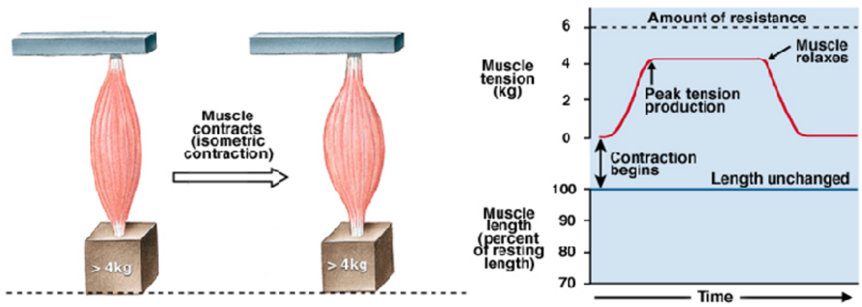
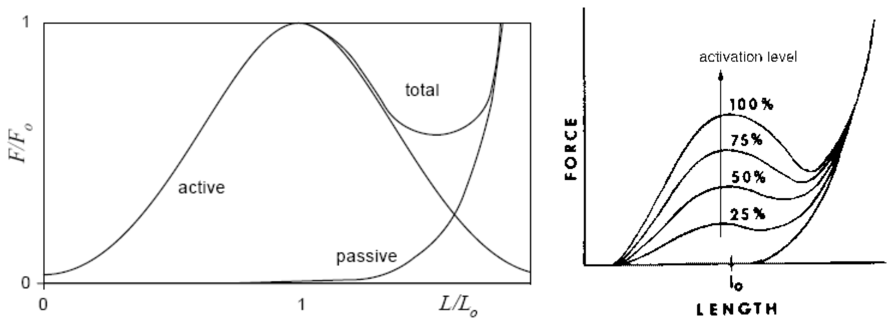


Figure 2.13: Isometric tension

One may expect that finding the slope of the active force-length curve would give a measure of stiffness of the muscle fiber. However, this plot is made out of data points, no length change is imposed during this measurement of force and also this seady-state force at a given length depends on the mechanical history of the muscle fiber, the slope of the steady-state length tension curve may not approximate the stiffnes accurately [22]. Stiffness due to steady state tension is called *static stiffness* (page 3.1).

Gordon *et.al.* in the 1960s explained that the *dome shape* is due to the changes of the structure of the myofibrils at the sarcomere level [20]. Without going into much detail in sarcomere level, from this active tension-length curve it is known that there is an *optimum muscle length* in which muscle create optimum active tension (Figure 2.14)a. Since it is the voluntary activated tension, the active tension-length relationship is also a function of activation. Thus, one muscle have many different force-length curve depending on the activation level (Figure 2.14)b.

The total tension-length relationship of a skeletal muscle is sum of the active and passive behavior (Figure 2.14)a. In fact, due to the difficulty of measuring the active behavior, the active tension is measured by subtracting the passive from the total tension of the muscle [34].



(a) F_o: optimum force L_o: optimum length (b) activation effect

Figure 2.14: Theoretical force-length curve of active element, passive element, and the total value

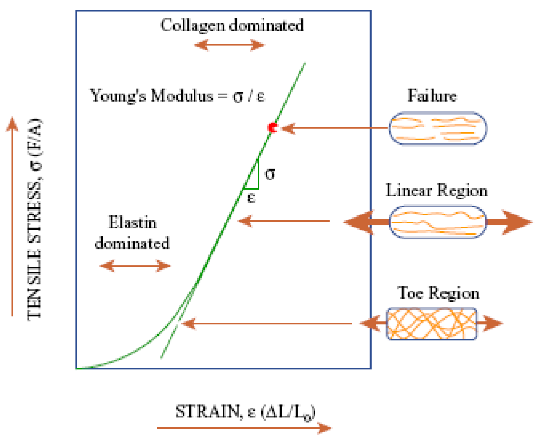


Figure 2.15: Stress-strain curve of collagen dominated tissue (MIT Online Courseware)

Unlike the active tension-length curve, the passive tension-length curve is obtained when muscle is stretched in absence of activation or active stimulation

[34]. As shown in Figure 2.14b, only the active component varies with activation level. The curve belongs to the passive element remains the same. The passive element contribute to the force-length relationship as the stretching is above optimum length or resting length (Figure 2.14)a. The passive element force-length curve has toe region (concave region) where force increases slowly with the stretching i.e non linear region (Figure 2.15). As the muscle is further stretched, force is linearly increased with strain until tendon failure (8-10% strain) (Figure 2.15). However, tendon tends to always operate at toe region even at maximum contraction [22].

Also, it is observed that relative peak extension of series element is about an order of magnitude less than that of the parallel element (Figure 2.16). In other word, SE is usually much stiffer (it creates maximum passive force at smaller strain) than PE over the primary operating range for most muscles [38].

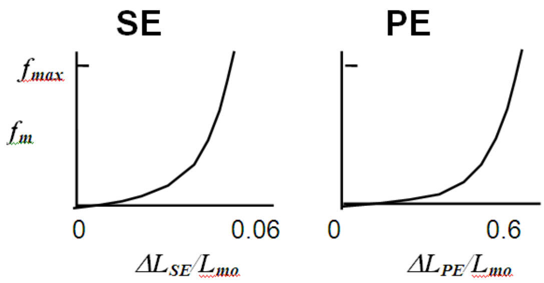


Figure 2.16: force-length curve of passive element [38] with x-axis as the stretch ratio (change of length/optimum muscle length)

It is suggested from in vivo experiments that the elastic modulus (E) of tendon varies from 0.45-1.2 GPa [22]. The tendon stiffness can also be written as followed,

$$K_{tendon} = \frac{AE}{l} \tag{2.3}$$

In other word, tendon stiffness increases with cross-sectional area (A) or thickness and decreases with length. Thicker tendons are stiffer and longer tendons are more compliant [22]. .

Constitutive Relationship: Isotonic Active Force-Velocity

Isotonic means constant force. A skeletal muscle may contract to the different length while maintaining its force (Figure 2.17). As seen in Figure 2.17, there is a drop in muscle length as time passes while maintaining the force constant under maximal neural drive i.e. sustained tetanus [38]. In other word, there is a change of length with time i.e. velocity. This velocity is observed almost constant. Therefore, repeating the experiment would give isotonic force - velocity data points that give the Force-velocity curve.

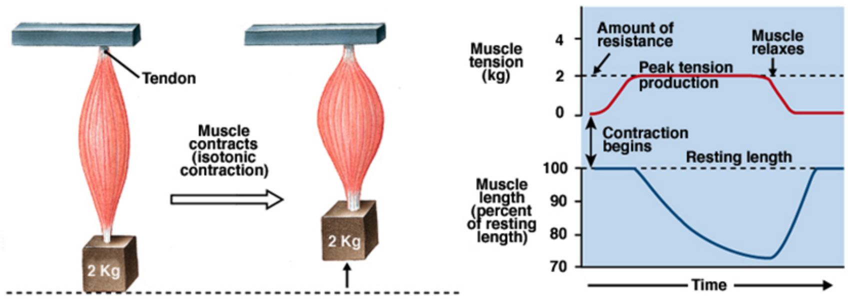
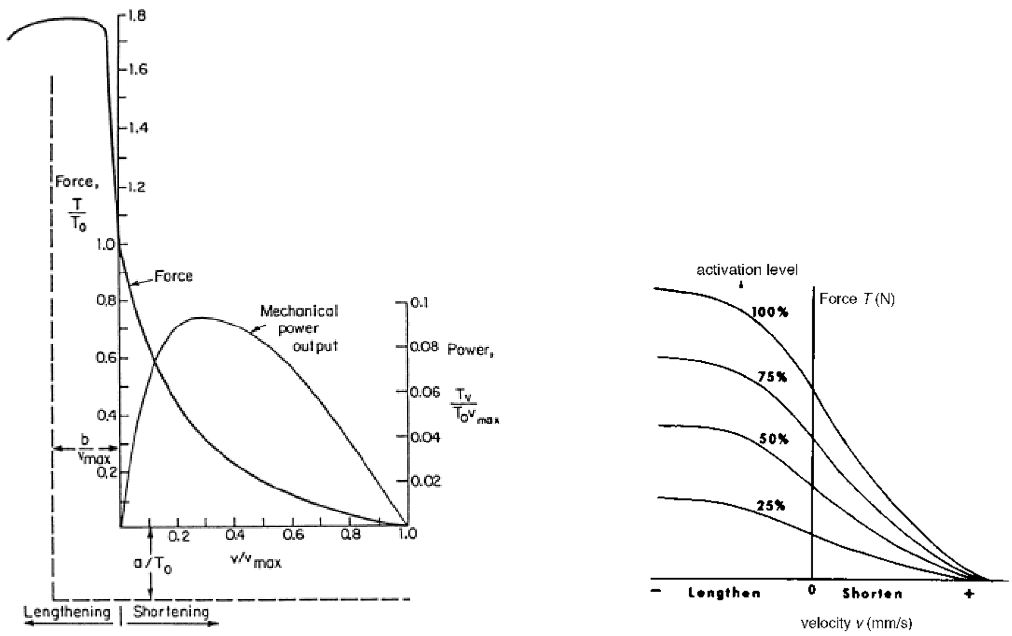


Figure 2.17: Isotonic tension



(a) Force-velocity curve and also the integrated power

(b) Effect of activation in the force-velocity characteristics

Figure 2.18: Force-velocity curve with T: Force

The force-velocity plot can interestingly be fitted into a square-hyperbola function [20], derived the first time by A.V. Hill as shown in equation 2.4 [38] and also shown in (Figure 2.18)a. Winters and Stark in 1985 modified Hill's equation through mathematical manipulation and obtained equation 2.5 [38].

The classical Hill's force-velocity:

$$(f + a_f f_h)(v + a_f v_{max}) = a f_h v_{max} (l + a_f) \tag{2.4}$$

After mathematical manipulation, equation 2.4 can be written as followed:

$$f_{ce} = f_h - f_b = f_h - [f_h + a_f / v_{ce} + a_f v_{max}] \, v_{ce} \tag{2.5}$$

Where,

f_{ce} : force of contractile element

f_h : hypothetical force, that is the "as if isometric" force crossing the vertical axis or at zero velocity

f_b : viscous force or velocity-dependent force

a_f : dimensionless constant; Hill's constant is usually 0.25

v_{max} : maximum velocity, that is the velocity crossing the horizontal axis or *unloaded maximum velocity*

In short, the force-velocity relationship is characterized by a rapid drop in the muscle force with increasing the shortening velocity (the right quadrant of figure 2.18a) and a rapid rise in the muscle force when the muscle is forced to lengthen (the left quadrant) [20].

A force-velocity graph as shown in Figure 2.18 can describe the *mechanical power output* that an active muscle delivers . Power is force times velocity. In a plot of force as a function of velocity, the power is the integration of force with respect to velocity (known in basic mechanics study and also stated in [38]).

From both figure 2.14b and figure 2.18b, the general effect of increasing the muscle activation level is the increase of muscle force (isometric tension and isotonic tension). The plot itself becomes steeper in the ascending and descending portion. Personally, this may be related to the increase of "stiffness" as a result of increase in activation. In dealing with instantaneous stiffness in complex movement, the focus of this study, the significance of these constitutive relationships should be considered and applied wisely because the plots correspond to somewhat discrete data points obtained from isometric and isotonic tests.

Effect of muscle architecture in constitutive relationship

The physiological cross sectional area affects the maximum force exerted by a muscle but not on the velocity. The larger the PCSA, the larger is the muscle force (figure 2.19). Eventually, one could also relate the effect of pennation angle to the maximum force the muscle can create. Longer fiber contract at higher speed for the given muscle force than the shorter fibers (figure 2.20).

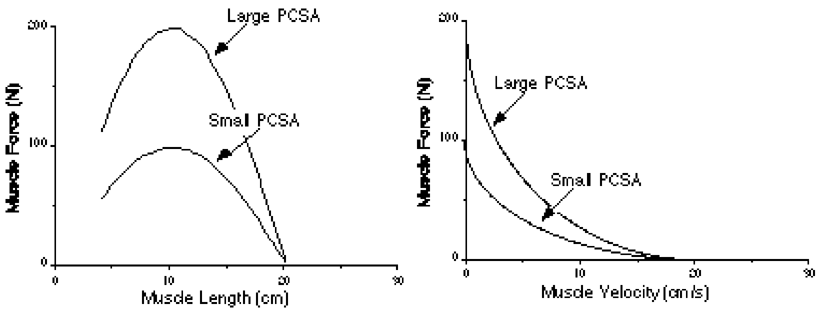


Figure 2.19: Effect of PCSA in force-length and force-velocity curves [4]

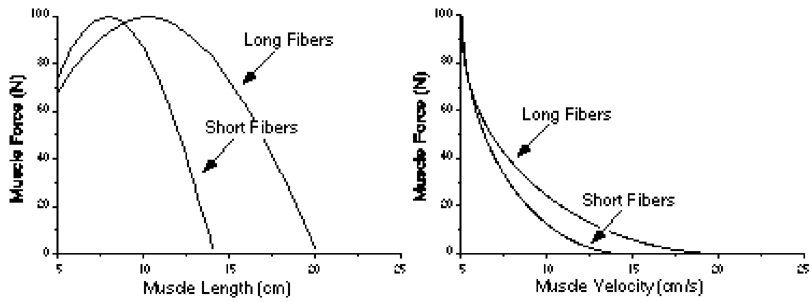


Figure 2.20: Effect of fiber length in force-length and force-velocity curves [4]

When the effect of both PCSA and fiber length is put together, the force length and force velocity curves are altered as shown in 2.21. It has been mentioned that the muscles, of which function is to produce large force or tension, have commonly large cross section and short fiber. However, short fiber means slower delivery of the force (contraction takes more time). On contrary, longer fiber that is able to contract faster and is usually associated to smaller PCSA, which means less intensity of force is produced.

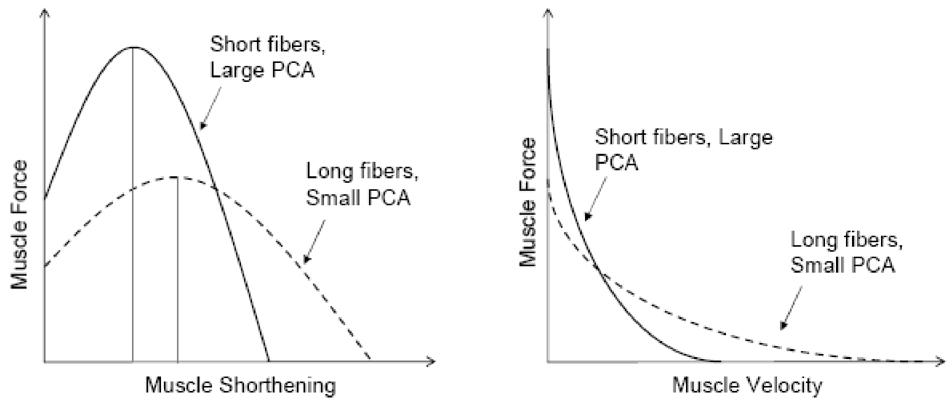


Figure 2.21: Effect of both PCSA and fiber length in force-length and force-velocity curves [4]

Chapter 3

Arm Stiffnes: Muscle, Joint and End-Point Stiffness

Studying the limb stiffness has a high importance in biomechanics. The biomechanists studying limb stiffness are motivated by the expected roles of stiffness in postural and movement tasks. An example in biomechanical sport application is presented in [13] discussing the control of stiffness in hoping that can be related to runners' performance. In biomedical application, [12] mentions that increase in stiffness results in less injury (i.e. lesion) in human muscles. In the study of stability, stiffness is also known to affect the movement control in reacting to perturbations [19]. Also, [19] presents the work studying muscle stiffness on pre-pubertal children. The effect of age in the stiffness of our body is common in daily life. It is particularly known that *passive joint stiffness* is higher in older population, people having rheumatic disease, most people in the morning, and people after undergoing surgery [40].

An interesting point mentioned in [12] is the role of *passive stiffness*, as a property of a muscular tissue, in setting the ability of the tissue to store energy. From solid mechanics, we know that the area under the curve in stress-strain plot is the energy absorbed by the body when elongated, in rest, or compressed and that the slope of the curve is termed stiffness. Thus, muscles with higher passive stiffness are more able to absorb energy.

3.1 Definition of Stiffness in Biomechanics of Limb

Defining the term 'stiffness' of the limb is rather not straight forward. In the study of mechanics, stiffness is defined as the resistance of an elastic body to deflection or deformation due to an applied force. In other word, stiffness equals to the amount of force needed to extend the body in one unit length, with SI unit is N/m [40]. Compliance is just a term to define the inverse of stiffness. Both terms, stiffness and compliance, are used widely in biomechanics dealing with deformable and elasticity property of our body. The followings are the terminology used in discussing limb stiffness:

- Based on the components of the limb, there are **muscle stiffness**, **joint stiffness**, and **end-point stiffness**. Joint stiffness is the amount of torque (Nm) increment per unit of joint angular deflection (radian or degree) [40]. End-point stiffness is the stiffness at limb's extremeties, i.e. hand, fingers, feet, head. Muscle stiffness include the stiffness of muscle fibers and the connective tissue.

- Based on the dependency on time, there are **static stiffness** and **dynamic stiffness**. The static stiffness is defined as the difference of steady state force at two different lengths. The dynamic stiffness of a muscle is the instantaneous stiffness measured during a change in muscle length. Static stiffness of a muscle is determined by measuring the difference in force between two steady states at different lengths, where the length is not changing. Dynamic stiffness, on the other hand, is the derivative of force with respect to length, which is defined only when length is changing. In measuring dynamic stiffness, the velocity-dependent and acceleration-dependent forces must be estimated and subtracted if they are not negligible [22].

- There are a few terms based on the fact that passive objects like spring need external force to deform, otherwise remains, and active objects are those that can change length or angle without external force. These terms are **passive stiffness**, **active stiffness**, and **apparent stiffness**. When an object deforms due to the applied force, like a spring, the resistance is addressed as passive stiffness. However, when the stiffness is due to an active force creation i.e. muscle contraction, it is called active stiffness. In active objects, stiffness is dependent on specific motor-task and time course [40]. It is also important to note, that there is no one-to-one relationship between the force or torque and the

> displacement in active objects [40]. In this case, connective tissue has passive stiffness. [40] introduces another term called apparent stiffness to address the stiffness-like behavior of active objects. An apparent stiffness what later be defined as second derivative of the spring-like potential energy [30] taking spring-like forces and constant forces in consideration. Muscle fibers, joint, and end-point have apparent stiffness.

It is also found in [12], the distinction between the terms *stiffness* and *flexibility* of muscle, two terms perceived as synonyms in our daily language. However, [12] clarifies the definition of muscle stiffness that is one of the mechanical properties of muscle related to the resistance of muscle to deformation and mentions that graphically, stiffness is represented by the stress-strain curve. Flexibility is the ability of the muscle to extend in all of its length breadth. In point of view of the joint, flexible joint can move in the whole range of motion.

In this thesis, the term muscle stiffness, joint stiffness and end-point stiffness are used. Since the instantaneous stiffness value is the input for HASy, the dynamic value of stiffness is the focus. Also, a muscle stiffness comprises of both active muscle fiber stiffness and passive connective tissue stiffness. The joint stiffness is an apparent stiffness because it is actuated by the muscles in postural and movement task.

3.2 Calculation of limb stiffness

Our musculoskeletal system is analogous to a multi-body multi-link system in mechanics with skeletal muscle as the actuators that cause movement in the system. The stiffness parameter of a musculoskeletal system is quantified when biomechanical researchers and modelers want to analyze the dynamics of the system, understand the fundamentals of equilibrium in human posture and movement, and the stability of this equilibrium. There are two types of stability, namely naturally stable and naturally unstable [40]. *Naturally stable* (*unintended stable*) system is when there are only external forces acting resulting in a stable system. Our musculoskeletal system, as a *kinematic chain* or *link*, is considered as *naturally unstable* (*intended unstable*) system[40].

The principle of stability or balance has been introduced in sub chapter 1.2.2. To start the discussion of limb stiffness, the concept of stability is further discussed here. According to [40], the equilibrium of a kinematic chain is stable when the followings are fulfilled:

1. **The forces and couples acting on the kinematic chain tend to decrease the position deviation**. In our musculoskeletal system, which is naturally unstable in most cases, stabilization, the effort to ensure that the forces and couples decrease deviation from equilibrium, is achieved mainly by *agonist - antagonist* muscle action [40].

2. **These acting forces and couples do negative work**. Negative work is done by a force, of which point of force application is moving against the force direction.

3. **The potential energy of the chain increases**. A muscle is mentioned to have a spring-like behavior. Analogous to spring, muscle has also the elastic potential energy that may increase as it stretched. Van der Helm also mentioned in [30] that a system is *stable quasi-statically* if the potential energy increases for *all* possible perturbations of the DOF. There, he also mentions that the potential energy in a musculoskeletal system is in two forms, each further categorized with respect to the location (intrinsic or external) as shown in Figure 3.1:

   - due to conservative forces
     - due to gravitational forces of body segmental mass (F_g)
     - due to constant forces applied at end-point (F_{ve})
   - due to strain in spring-like elements
     - of muscles (F_m)
     - of passive structures (F_p)
     - of environmental structure (F_{se})

It is important to note that in musculoskeletal system there are these five sources of potential energy, whether or not the system ends in stable equilibrium. Most importantly, there is a stiffness value contributing to the spring-like potential energy. The *apparent stiffness* is no other than the 2^{nd} *derivative of the potential energy.* Also, the fact that our limb is a musculoskeletal system, which is a kinematic chain interacting with the environment, imposes the complexity and interrelationship of stiffness values in our limb. In other words, in addition to the stiffness as an intrinsic property, the external effect

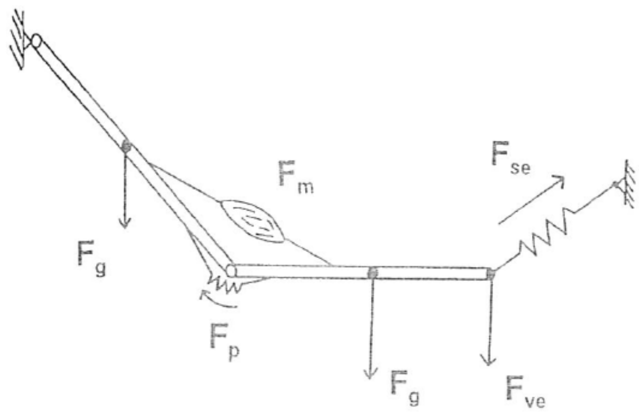


Figure 3.1: Potential energy sources in simple arm. The system will find an equilibrium for any combination of weight (conservative) and spring constant (strain-based) [30]

and also stability criteria should not be neglected in the real scenario. In depth formulation of potential energy, which is related to the stability requirement for quasistatic case, is outside the scope of this thesis but can be found in detail in [30].

The complexity of the problem is lifted in case of dynamics. In dynamic stability analysis, the proprioceptive feedback is taken into account [30]. It is suggested that when energy expenditure in muscles should be less and that time delay is not a problem, the reflex feedback gain is an optimum way to increase stiffness. Including feedback gains in a musculoskeletal model would affect the effective stiffness. Stability analysis of non-linear dynamic system (e.g. Linearization, Lyapunov approach) has not been available in *any* musculoskeletal model [30]. As the future direction of realistic large-scale musculoskeletal model is suggested to include stability requirement (quasi-static and dynamic) [30], [32], the respective significance in stiffness analysis is worth mentioning here.

3.2.1 Muscle stiffness

The concept of quasi-static stability has introduced the idea of spring-like source of potential energy in the musculoskeletal system. It is stated that the spring-like behavior of muscle is originating from several sources, namely the cross-bridge dynamics and reflex feedback (Section 4.4 of [30]). According to Van der Helm in [30], the followings hold for muscle stiffness:

- When the amplitudes of perturbation is small (a few milimeters), the stiffness of cross-bridge results in the muscle stiffness. It is mentioned by Kirsch and Stein in [17], that this is called *transient component of muscle elasticity*. This stiffness component, originated from the cross-bridges, can be substantial (larger than the slope of the force-length relation) but does not persist indefinitely. For small enough displacement, this stiffness has been shown as purely elastic that is without dynamics and proportional to displacement.

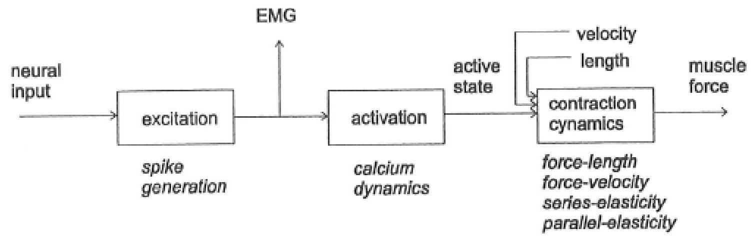
- For larger amplitude of perturbation, the *derivative of the muscle force-length relation* is equivalent to the muscle stiffness. This is the so called *steady-state component* of muscle elasticity [17]. According to Kirsch and Stein, this stiffness is rather modest but the one that provides a maintained response to a maintained disturbance. Also, It is known that above optimum length, the slope is negative. In other words, the equivalent muscle stiffness is negative and in the absence of passive stiffness, instability may occur. The importance of passive component is the instantaneous force response to disturbances, mostly important in postural stability maintenance in the presence of unexpected disturbances, which cannot be done by delayed neural reflexes [17].

The muscle dynamics applied in DSEM's forward dynamics simulation is the one developed by Winters and Stark [29]. It is a 3^{rd} order muscle model including a *Gaussian-shaped* active length-tension curve and non-linear series-elastic (SE) and parallel-elastic (PE) elements. Winters and Stark in [39] define the active length-tension curve as *torque-angle curve* and state that a *gaussian-type equation with linear "sloping" parameter* (figure 3.2b) is used because it is most adequate for the entire range of joints. The model provides a *nearly linear force-stiffness relation* where SE is much stiffer than CE, except at very low activations, and PE is much more compliant, except at high lengths [32]. Torque-angle equations for the active contractile element, series element and parallel element are presented in Appendix 1(73). Detailed discussion of the model can be found in the original paper [39].

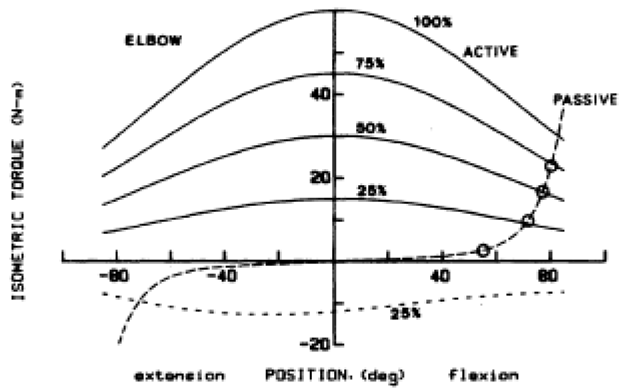
The model of Winters and Stark is Hill based (Figure 2.8 left, in sub chapter 2.4.1). It is well understood that the mechanical impedance of elastic elements in series sum inversely and the elastic elements in parallel simply sum their impedance [22], [32]. Thus, the stiffness of a single muscle consisting of CE, SE, and PE is,

$$k_m = \frac{k_{ce}k_{se}}{k_{ce} + k_{se}} + k_{pe} \tag{3.1}$$

Now, since SE is commonly much stiffer than CE, $\frac{1}{k_{se}}$->0. Also, as PE is much more compliant (lower stiffness),k_{pe}->0. In the end we are left with k_{ce} only. Therefore, k_{ce}, which is the slope of the active tension-length relationship, can be the first approximation of muscle stiffness. This is not true in the whole range of muscle action that varies with activation level, length and velocity. Also, this analysis does not take reflex feedback into consideration. As mentioned, however, the reflex feedback in musculoskeletal should be taken into account as it affects the effective stiffness.



(a) Muscle model block diagram [30]



(b) Active torque-angle [39], solid lines are elbow flexor torque-angle fit with activation level.

Figure 3.2: Muscle dynamics based based on Winters and Stark work (1985)

The block diagram shown in figure 3.2a has an excitation dynamics block, an activation dynamic block, and contraction dynamic block. Excitation block is a linear 1^{st} order model acting as a filter of neural input. Activation block is a non-linear 1^{st} order model describing the fast increase in calcium concentration and the slow decrease by the calcium pump. The contraction dynamics model has been described earlier. It has a contractile element, series elastic element and parallel elastic element. This dynamics used in the forward muscle dynamics simulation of DSEM.

Knowing that there are three different components contributing to the muscle stiffness value, one can derive each by differentiating the torque-angle equations. First, differentiating the torque-angle relationship (shown in equation A.1 and A.2, page 73) to estimate the k_{ce},

$$\frac{dMh\left(q\right)}{dq} = Nea \cdot \frac{dMXrat\left(q\right)}{dq} \cdot Mmax \tag{3.2}$$

Where $\frac{dMXrat(q)}{dq}$ can be derived using *the chain rule*,

$$\frac{dMXrat\left(q\right)}{dq} = exp\left[-\left(\frac{q - MXoo}{MXsh}\right)^2\right] \cdot \left(-\left(\frac{2 \cdot q - 2MXoo}{MXsh^2}\right)\right) + MXsl \tag{3.3}$$

Now, differentiating the passive torque-angle relationship of parallel element (PE) to get k_{pe},

$$\frac{dM_{pe}\left(q\right)}{dq} = ko + bo\frac{dv}{dq} + k1_{pe}\left(k2_{pe} \cdot exp\left[k2_{pe} \cdot q\right] - 1\right) \tag{3.4}$$

And k_{se} from the passive torque-angle relationship of series element (SE),

$$\frac{dM_{se}\left(q\right)}{dq} = ko + k1_{se}\left(k2_{se} \cdot exp\left[k2_{se} \cdot x\right] - 1\right) \tag{3.5}$$

Complete description of the variables and constants can be found in Appendix 1 (page 73). Those equations from Winters and Stark are the resulted joint torque (M)-angle(q) relationships due to the three lumped elements of the muscle (CE, SE, and PE). In the original paper, [39], the angle is addressed with variable-x.

Now, putting the torque-angle relationship in terms of muscle force and muscle length, one can generally write the followings,

Because,

$$dM = dF_m \cdot r_{arm} \quad ; \quad dl_m = r_{arm} \cdot d(q) \tag{3.6}$$

Then the relationship between torque-angle slope and the force-length slope is

$$\frac{dM}{d(q)} = \frac{dF_m}{dl_m} \cdot r_{arm}^2 \tag{3.7}$$

It can be seen from equation 3.7 that muscle stiffness scales with the square of the muscle moment arm and from equation 3.6, the muscle force scales directly with the moment arm (as also pointed out in [32]). It is stated in [32] that the latter can be larger than the former one, and the net contribution of the muscle to the joint stiffness may be negative [32]. Stiffness times the square of the moment arm (equation 3.7) is always positive for positive muscle stiffness. When the force magnitude is constant, the joint stiffness is contributed only by the change of moment arms. For example, when viewing human skeleton as inverted pendulum, gravity forces destabilizing [32] as it is often the case.

More about the factors affecting muscle stiffness, it is stated that the combined stiffness, viscosity, and inertial relations characterize the *mechanical impedance*, which is tunable based on muscle activation [24]. Muscle stiffness depends on the force exerted by the muscle as shown from experimental results that the larger the muscle force, the greater the resistance to the stretch provided by the muscle [40].

In *in vivo* experiments, Muscle stiffness is intricate to calculate. Because relation between length and force depends on mechanical history of muscle, the calculated stiffness depends on the stiffness measurement method. The stiffness of the muscle fiber, specifically, depends on the cross-bridges attachment. In results, the measurement of instantaneous stiffness is valid when the variation of the three parameters (activation level, length and velocity) is small enough during the measurement interval that the number of attached cross-bridges do not significantly change [22], [32]. Two methods that are commonly used to estimate the *instantaneous* stiffness of muscle fiber are as followed [22]:

- Imposing a muscle sample with rapid, but constant velocity, small length change so that the velocity dependent forces are negligible in comparison to elastic force. From this experiment, instantaneous stiffness of muscle fiber is estimated from the slope of the force plot against length.

- Imposing high frequency (1-4 kHz) sinusoidal and small (0.1% sarcomere length) length changes. Here, the force at peak displacement is measured, when the velocity-dependent forces are negligible as velocity is zero. Even if acceleration at this time is maximum and acceleration-induced force can be calculated by multiplying the muscle fiber mass with the acceleration, the mass of muscle fiber is so small that this can be neglected.

It can be seen that these methods put the condition in such a way that the velocity-dependent and acceleration-dependent forces can be neglected. In other word, viscous and inertia effect is avoided.

3.2.2 Joint stiffness

Commonly, to analyze the equilibrium of naturally unstable kinematic chains, the joint stiffness or joint compliance is examined. According to [40], joint stiffness matrix includes both direct terms and cross-coupling terms. Direct terms relate changes in the torque at a given joint to the angular deflection at this specific joint. Cross coupling terms relate changes in the torque at a joint to angular displacements of another joint.

It is known that joint stiffness can be controlled by co-contraction of the antagonistic muscles around the respective joint [38]. This can result in increase of joint stiffness without changing the joint position. Limb is a kinematic chain and there is a mathematical mapping from skeletal muscles to joints. The reverse may not be possible mathematically.the joint stiffness could be calculated using a mathematical mapping from the known muscle stiffness. The following summarizes the basic of the mapping as explained in [40].

First, letting the stiffness of i-th muscle be,

$$K_{m,i} = \frac{dF_i}{dl_i} \tag{3.8}$$

Where $\mathbf{K}_m$ is the the muscle stiffness matrix, of which dimension is mxm, and m is the number of muscle . Also, muscle length can be represented as a vector, which is a function of q, which is the joint angle or coordinates or degrees of freedom (DOF).

$$\mathbf{L} = f(q) \tag{3.9}$$

Now, assuming infinitesimal displacement around q is considered, the following holds:

$$d\mathbf{L} = \mathbf{J}_{mj}(q)dq \tag{3.10}$$

$\mathbf{J}_{mj}$ is the *muscle Jacobian matrix* consisting partial derivatives that relate infinitesimal changes of muscle length with the infinitesimal changes of joint angle. This mapping is also known as *geometric compatibility* mapping. Equation 3.10 is exactly the same as the equation 2.1 in section 2.3 when discussing about the moment arm, except that d$\mathbf{L}$ was dx_m and dq was dx_q. In other word, muscle jacobian matrix is no other than the muscle moment arm matrix. Moment arm can be calculated by measuring the muscle length that corresponds to a certain joint displacement (equation 3.7 right) [38].

$$\mathbf{J}_{mj} = \begin{pmatrix} \frac{\partial L_1}{\partial q_1} & \frac{\partial L_1}{\partial q_2} & . & \frac{\partial L_1}{\partial q_k} \\ \frac{\partial L_2}{\partial q_1} & . & . & . \\ . & . & . & . \\ \frac{\partial L_m}{\partial q_1} & . & . & \frac{\partial L_m}{\partial q_k} \end{pmatrix} \tag{3.11}$$

Letting F_m be the 1 x m matrix of muscle force, the joint torque $\mathbf{M}$ is as followed,

$$\mathbf{M} = \mathbf{J}_{mj}^T \cdot F_m \tag{3.12}$$

Where, $\mathbf{J}_{mj}^T$ is the transposed muscle Jacobian matrix.

The joint stiffness matrix can then be calculated from the muscle stiffness matrix as followed,

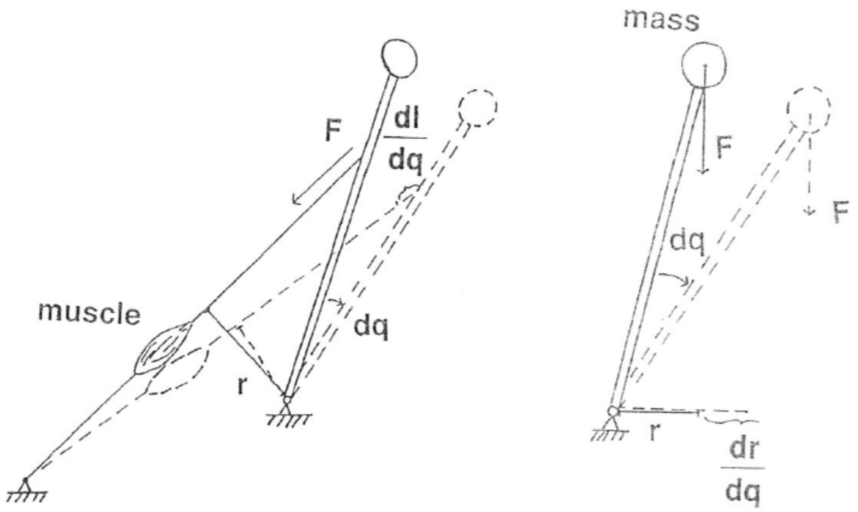
$$\mathbf{K}_j = \mathbf{J}_{mj}^T \cdot \mathbf{K}_m \cdot \mathbf{J}_{mj} \tag{3.13}$$

The more general apparent joint stiffness includes the stiffness-based, the term shown in equation 3.13, and also the conservative force-based [38]. The apparent stiffness of joint is,

$$\mathbf{K}_j^{apparent} = \mathbf{J}_{mj}^T \cdot \mathbf{K}_m \cdot \mathbf{J}_{mj} - F_m \frac{\partial J_{mj}}{\partial q} \tag{3.14}$$

Apparent joint stiffness is related to the stability of the system.

$K_j^{apparent} = \frac{\partial^2 E_{pe}}{\partial q^2}$ must be positive definite that is the eigenvalues of this symmetric matrix must all be positive for the system to be stable [38]. In [38], the joint coordinate q is addressed as x_j. The concept of total potential energy (E_{pe}) has been introduced in the begining of this sub chapter.



(a) Joint stiffness induced by the muscle spring-like behavior

(b) Joint stiffness from the conservative force

Figure 3.3: The apparent joint stiffness is contributed from the conservative force and spring-like force [30]

It is important to recall that our musculoskeletal system is an actuator-redundant system that there are a lot more muscles available than the joint DOF. Thus, there is an infinite possible solution of muscle forces to the same joint displacement problem. This is called the load sharing problem in musculoskeletal modelling. Due to the *actuator redundancy*, the inverse problem that is to find muscle stiffness from the joint stiffness in modelling a musculoskeletal system requires the inclusion of optimization criterion. Also, there are basically two strategies to increase the stiffness of a joint, namely by co-contraction of antagonistic muscles and defining the musculoskeletal system as a control loop, increasing the gains of the feedback loops.

Finally, like muscle, in vivo measurement of the joint stiffness may not accurately be defined simply by the ratio of joint torque to angular displacement without care taken in the effect of velocity and acceleration [40]. To summarize, one can analyze the dynamics of the joint by modelling the joint as a viscoelastic hinge where resistance is provided by both elastic and damping forces and that the angular motion as rectilinear deformation. Thus, the joint force can be defined as,

$$\mathbf{F}_q(t) = m\ddot{x} + b\dot{x} + k(x - x_0) \tag{3.15}$$

Where,

$\mathbf{F}_q$: Joint force with q-DOFs of the joint

m : mass, the linear measure of inertia

b : damping constant

k : spring constant

x : rectilinear position vector representing the joint displacement

$\ddot{x}$: acceleration, $\frac{d^2x}{dt^2}$

$\dot{x}$: velocity, $\frac{dx}{dt}$

Then deriving the equation with respect to x we get,

$$\frac{d\mathbf{F}_q}{dx} = m\frac{\dddot{x}}{\dot{x}} + b\frac{\ddot{x}}{\dot{x}} + k \tag{3.16}$$

Where $\dddot{x}$ is called *jerk*. Here, it is shown clearly that deriving the force with respect to to the displacement, $\frac{d\mathbf{F}_q}{dx}$, is equal to spring constant, k, only when the acceleration and velocity is zero and that the measurements are performed somewhat at equilibria. Otherwise, $\frac{d\mathbf{F}_q}{dx}$ would actually include the effect of inertia i.e. mass, viscosity (b), and spring constant (k). This means that when the subject is moving, simply deriving the force with respect to displacement may not be accurate. Instead, the term *mechanical impedance* should be used.

3.2.3 End-point stiffness

The endpoint behavior is commonly viewed important because that is where most mechanical interactions take place [22] and it is perhaps understood the best in comparison to the other stiffness value. It might be the easiest stiffness value to be estimated in experiments. It is also realized that the term *arm stiffness* is tightly related to the work of Mussa and Ivaldi in 1985 that formalize the arm stiffness geometrically, which is the *stiffness ellipse*

[23]. Note that this stiffness ellipse corresponds only to the end-point or end-effector i.e. the hand. The end-point stiffness is a non-scalar quantity and is usually defined in cartesian coordinates.

An interesting fact of end-point stiffness is that it is a non-scalar value. It is represented by a matrix and alternatively, by a stiffness ellipse. The derivation of the stiffness ellipse can be found in [23]. This is the 2D planar case. The significance of this geometrical representation, the stiffness ellipse, is shown in figure 3.4:

- The ellipse shows the magnitude of the generated restoring force in response to a unit magnitude of deflection from the equilibrium position [40]. The major axis of the ellipse is the maximum stiffness axis and the minor axis is the minimum stiffness axis. Also, only along these major and minor axis that the restoring force is in-line with the displacement.

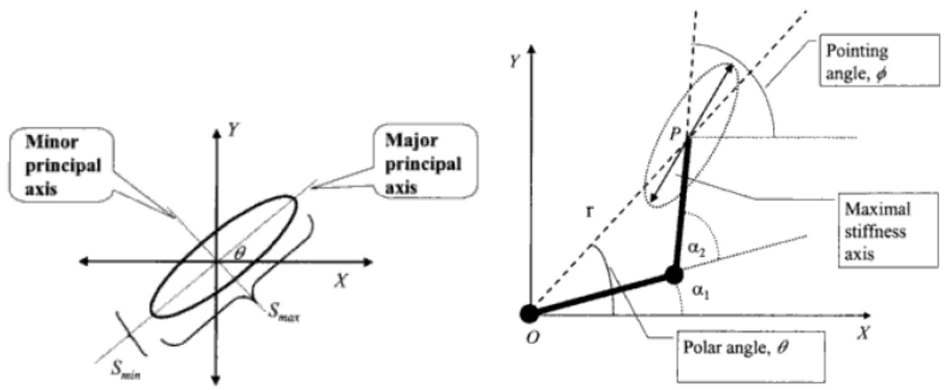
- The size of apparent stiffness ellipse varies among different subjects i.e. person and is not reproducible in measurements at different times i.e. days [23],[40].

- The shape and orientation of the stiffness ellipse depend largely on the arm configuration [40]:
  - The major principal axis intersects or passes close to the shoulder joint
  - When arm is extended and hand moves further from the shoulder, the ellipse becomes more elongated
  - In contrary, when the hand approaches more proximal location (closer to the body), the ellipse becomes circle-like or more isotropic.

More important is the discussion of the mathematical mapping because it is applicable also in 3D case. The mathematical mapping between the joint and the end-point stiffness is called the *chain Jacobian matrix*. Chain jacobian matrix relates the small change of joint displacement to that of the end-point. The relationship is as follows,

$$\mathbf{K}_j = J_{ej}^T \mathbf{K}_e J_{ej} \tag{3.17}$$

Special case is the planar 2-joint i.e. shoulder and elbow model where one could say that if the end-point stiffness (K_e) is stable and the joint stiffness



(a) Stiffness ellipse showing the major and minor axes (b) Planar two-link model of human arm

Figure 3.4: The perimeter of the ellipse is the locus of the force vectors for unit displacement in $0 \leq \phi \leq 2\ \pi$, from [40]

(K_j) is also stable. Here one could also calculate the reverse. Calculating 2 x 2 K_e in terms of 2 x 2 K_j with the following,

$$\mathbf{K}_e = J_{ej}^{-T} \mathbf{K}_j J_{ej}^{-1} \tag{3.18}$$

Although stability is not the focus in this study, the fact that the musculoskeletal system is a *kinematic redundant system* should not be forgotten. Due to this kinematic redundancy, it is not enough to consider only the endpoint stiffness in stability analysis. [38] says that there can be "hidden" joint instabilities.

3.3 Effect of Muscle Activation Level on Muscle Stiffness

The control of movement by the neuromuscular system requires coordination between the *physiological properties of the neural activation system* and the *biomechanical properties of the muscle-tendon unit* [36]. It is clear the muscle activation level plays an important role in defining the exerted muscle force and in results, the muscle stiffness. Commonly, activation level is used to scale the force-length relationship. The force-velocity relationship is usually scaled in such that only the maximum velocity is affected i.e. the corresponding force at zero velocity is constant. An example in real life is the effect of age to limb stiffness. In [19] imposes the necessity to take into account the capacities of muscle activation to quantify changes in elastic properties of muscles because *muscle activation is altered with age.*

Lower muscle activation levels are prevalent in everyday life [32]. It has been mentioned that at very low activation, SE i.e. tendon is much stiffer than CE and of course PE (3.2.1). In other words, tendon stretch plays a minor role in altering length-tension properties at low activation levels [26]. In reality, even at maximum contraction, tendon strain is still observed to be small or operates in the toe region [22]. Indeed, tendon must be sufficiently stiff in most muscle operation range so that it can function well (to transmit force to bones).Also, it should be noted when dealing with low excitation levels. differentiation of the length-tension relation is not sufficient to define *apparent muscle stiffness* [32]

From the gaussian active torque-angle curve, as activation level increases, the slope on either sides of the torque-optima also increase. In other words, the stiffness of active CE increases. [39] concludes that excitation and activation processes are most significant in *fast voluntary movements* while the torque-angle and the parallel elastic non-linearities dominate near *the movement-range extremes* and the series elastic element is important in tasks with *transient external loading.*

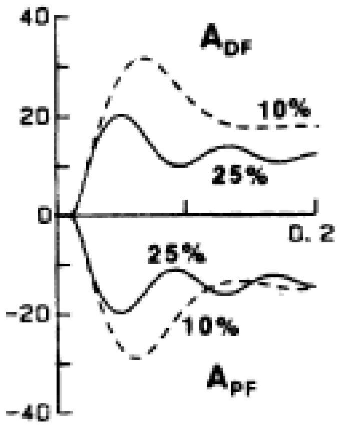


Figure 3.5: Effect of activation dynamics [39]

Figure 3.5 is the effect of 10 ms of 100 Nm external torque impulse disturbance on the ankle model, under two different cocontraction levels (taken from [39]). The responds show the open loop cocontraction effectively changes the system "stiffness," and the f*requency of oscillation increases with coactivation level.* It is seen that higher coactivation level (25%) imposes better impedance (or damping or resistance) but results in higher oscillations (or less stable respond). 10% coactivation level shows worse damping to the external impulse but less oscillations.

When activation or neural input is high, consequently the muscle force is high and there is not much change in the difference between total muscle displacement is almost equal to the CE displacement. Therefore, *SE could potentially be ignored or at least linearized over the operating range of interest* [38].

The followings are a view other interesting facts regarding the effect of activation level in stiffness value and production:

- It is found in [22] that the muscle-tendon (musculotendinous) unit acts like a *filter* whose bandwidth and gain depend on the muscle activation level. The higher the activation, the higher the stiffness due to increase in the gain of the filter [22].
- As the activation changes, the CE relation sculpts (shape) force, roughly instantaneously [38].
- [32] states that when muscle activation is coordinated, some muscles will be activated not only because of their contribution to the net joint moment, but also for their contribution to the joint stiffness. In other words, muscle activity shapes the endpoint stiffness field.

Finally, in space medicine study, especially of skeletal muscle, it is found that muscle activation level in microgravity is lowered; thus, the same level of muscle stiffness is more difficult to achieve for the same amount of muscle force requirement [10], [37]. The reduction of activation level is also observed after immobilisation and bed rest. The reduced activation level is observed in posture muscles. It is suggested that this is due to the reduced requirement to whitstand the weight and maintain stability due to the gravitation force.

Chapter 4

Arm Stiffness Using DSEM as a Biomechanical Model

Having understood the biomechanics of human musculoskeletal system, especially in dealing with stiffness of the upper limb, the calculation of muscle and joint stiffness could be added in a biomechanical model of human upper limb. DSEM is used as the biomechanical model in this project. The following figure shows schematically the use of the biomechanical model as a platform to provide reasonable value of joint stiffness for DLR HASY system.

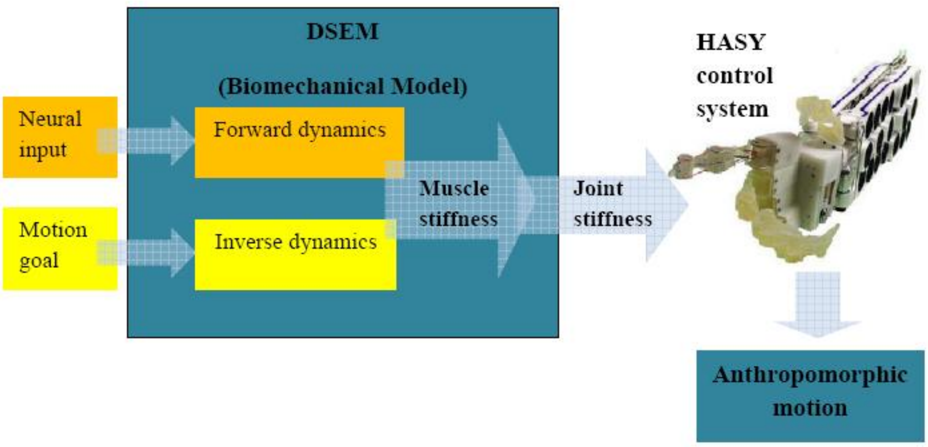


Figure 4.1: Flowchart showing relationship between DSEM and HASY

4.1 DSEM Introduction

Delft Shoulder and Elbow Model (DSEM) is the biomechanical model used in this project. DSEM is a 3D finite element model (FEM) of the human arm (shoulder-elbow complex). It was developed in Delft University by Van der Helm [28], [29]. The model is able to do both inverse and forward dynamic analysis. The definition of inverse and forward dynamics are presented in section 2.4.1. The model does the analysis making use many parameters like the muscle length, the respective muscle force, the joint rotation and orientation, and the joint torque (moment). From basic mechanics, stiffness is defined as the relation of a force applied on a deformable body with the resulted deformation. However, extracting the stiffness of muscle and joint in musculoskeletal system from the model is not as straight forward as looking at the original DSEM output data.

The development of this shoulder-elbow model has threefold goals, namely:

- To enhance the functional insight in the mechanics of the upper extremity, and the role of the various morphological structures
- To investigate some clinical questions about the diagnosis and treatment of shoulder and elbow complaints
- To calculate the loading of morphological structures during work, ADL (loaded abduction) and other activities, in order to relate the loading with persistent shoulder complaints. Subsequently, if it is revealed what is causing the high loads and/or what are the vulnerable structures, the tasks can be adjusted in order to prevent the complaints.

It can be seen that the originally aimed applications of DSEM are enhancement of biomechanical study and also medical. There are available papers that review the application of DSEM for medical analysis like the loading analysis for wheelchair user [33]. There are several other clinical applications like the sensitivity analyses of a shoulder arthrodesis and a glenohumeral endoprosthesis [38].

DSEM, a finite element model

There are several ways of modelling a complex multi body system. DSEM itself is developed as a finite element model of the shoulder-elbow complex, developed using SPACAR program. Using SPACAR, each component of the shoulder-elbow complex are represented by an appropriate element, of which the mechanical behavior is known [38]. The elements are connected through shared end nodes. In this way, the mechanical behavior of the whole system can be derived. The followings summarize the elements used to represent the components of the shoulder-elbow complex [28]:

- Bones are modeled as single, rigid BEAM elements and are multi node (MNODE) elements. This element has local coordinate systems and attachment sites fixed in the body.

- Scapula is modeled as combination of two rigid BEAM elements

- Three synovial joints, sternoclavicular, acromioclavicular and glenohumeral joints are each represented by 3 orthogonal HINGE elements.

- Muscles are modeled as one or more active TRUSS or CURVED-TRUSS elements that can lengthen and shorten to deliver the force necessary for position changes and dynamic equilibrium. CURVED-TRUSS is used to model muscle that wraps around a bony contour. In current DSEM, there are 31 muscles included with most of the muscles have a large attachment site and are poly-articular. In order to represent the mechanical effect of a muscle with large attachment site, more than one TRUSS or CURVED-TRUSS element is used. The 31 skeletal muscles in current DSEM are represented by 139 elements. Table 4.1 shows the number of elements representing the 30 arm muscles measured from a cadaver with one muscle without data from the cadaver measurement.

- The costoclavicular, conoid, trapezoid ligaments are modeled as flexible, passive TRUSS elements.

- The scapulothoracic gliding plane is modeled by two SURFACE elements, which define the compression forces between the medial border of the scapula and the thorax.

Mass and rotational inertia of segments are represented by lumped inertia at the nodes and external forces (e.g. gravity) also act at these nodes. The dynamical behavior is described in user-defined subroutines (in FORTRAN). The post-processing can be done in MATLAB (PlotDSM program) and also SIMM (Musculographics Inc.) as shown in 4.2. Figure 2.10 in section 2.4.1 show the finite element representation and SIMM visualization of DSM, which is the older version of DSEM.

As a finite element model, there is a set of variables defining position and orientation (in terms of coordinates) and the deformation of an element (deformation vector). For all elements in the system (musculoskeletal system of shoulder-elbow complex), the mathematical relation between the *deformation vector* $\boldsymbol{E}$ and the *position vector* $\boldsymbol{X}$ is calculated (Appendix 4 of [38]). New elements, such as the CURVED-TRUSS and SURFACE in DSEM, can be developed by defining a certain relation between the deformation of the element and its coordinates. The motion equations are derived for the whole mechanism using the *principle of virtual work* and *d'Alembert principle.*

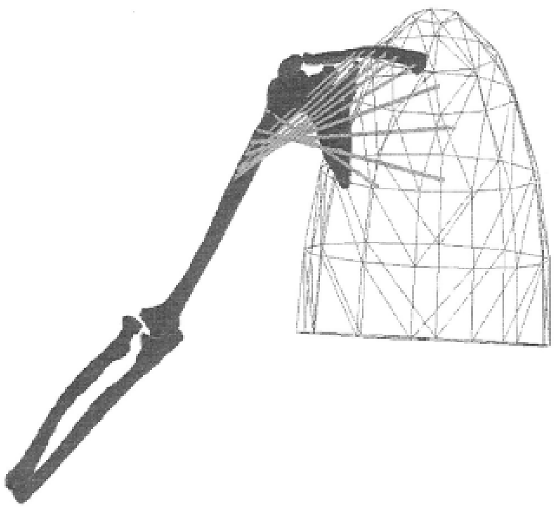


Figure 4.2: DSEM visualization with SIMM [38]

Before going in details about the dynamics simulations can be done with DSEM, the fact that a muscle might be represented by many elements with different spatial coordinates should be pointed out. In our interest in calculating muscle stiffness, one may expect that one muscle would have a single stiffness value at a specific time and neural activation. However, this may not be the case when deriving muscle stiffness from DSEM. In first approach, one could say that for large muscles with large attachment sites, there would be spatial definition of stiffness according to the spatial definition of the specific element. One could also obtain the resultant of the stiffness of each of this element to represent the stiffness of the muscle. However, the mechanical significance of this summation should be further studied.

Table 4.1: 30 arm muscles included in DSEM and the number of representing elements.

No.	Muscle name	Elements	Element number
1	M.TRAPEZIUS, SCAP. PART	11	1-11
2	M.TRAPEZIUS, CLAV. PART	2	12,13
3	M.LEVATOR SCAPULAE	2	14,15
4	M.PECTORALIS MINOR	4	16-19
5	M.RHOMBOIDEUS	5	20-24
6	M.SERRATUS ANTERIOR	12	25-36
7	M.DELTOIDEUS, SCAP. PART	11	37-47
8	M.DELTOIDEUS, CLAV. PART	4	48-51
9	M.CORACOBRACHIALIS	3	52,53,54
10	M.INFRASPINATUS	6	55-60
11	M.TERES MINOR	3	61,62,63
12	M.TERES MAJOR	4	64-67
13	M.SUPRASPINATUS	4	68-71
14	M.SUBSCAPULARIS	11	72-82
15	M.BICEPS, CAPUT LONGUM	1	83
16	M.BICEPS, CAPUT BREVE	2	84,85
17	M.TRICEPS, CAPUT LONGUM	4	86-89
18	M.LATISSIMUS DORSI	6	90-95
19	M.PECT. MAJOR, THOR. PART	6	96-101
20	M.PECT. MAJOR, CLAV. PART	2	102,103
21	M.BICEPS, CAP.LONG.ELBOW	1	104
22	M.TRICEPS, MEDIAL PART	5	105-109
23	M.BRACHIALIS	7	110-116
24	M.BRACHIORADIALIS	3	117,118,119
25	M.PRONATOR TERES, hum-rad	1	120
26	M.PRONATOR TERES, uln-rad	1	121
27	M.SUPINATOR, hum-rad	-	-
28	M.SUPINATOR, uln-rad	5	122-126
29	M.PRONATOR QUADRATUS	3	127,128,129
30	M.TRICEPS, LATERAL PART	5	130-134
31	M.ANCONEUS	5	135-139

Running a dynamic simulation with DSEM, the muscle length (**plen**), muscle length relative to initial, muscle length relative to optimum (**prellen**), and muscle force(**fmus**) are calculated for each element. This means, a muscle with many elements would also have as many output variables in each integration step.

4.2 Muscle Stiffness from the dynamics simulation

4.2.1 Inverse dynamics

Through inverse dynamics one could estimate the joint and muscle forces from the recorded motion input. Mathematically, as the position, time, velocity and acceleration of the pre-defined generalized coordinates are known, the system differential equations becomes a system of algebraic equations. The drawback of pure inverse dynamics simulation is that kinematic assumptions must be made. Thus, authenticity of the simulation depends largely on the chosen assumptions. The advantage is known widely that is the efficient optimization or computational effort. An example of the application of this simulation can be to understand shoulder pain for wheelchair users; thus, finding out shoulder joint loading [8]. The specific joint of interest here is the glenohumeral joint. Using DSEM and the PlotDSM the glenohumeral loading can be visualized and analyzed.

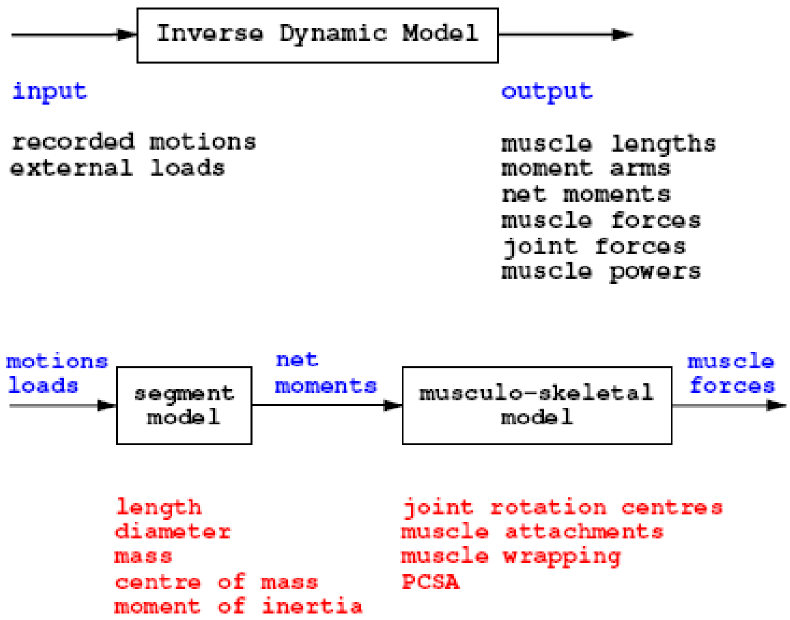


Figure 4.3: The inverse dynamics simulation in DSEM [8]

Unfortunately, as far as the interest in calculating the arm stiffness, especially the muscle stiffness, the inverse dynamics simulation program of DSEM is not sufficient. This is simply because there is no muscle dynamics in this simulation. Thus, neural excitation and activation corresponding to the motion can not be analyzed.

After studying the DSEM subroutines and defining all the output variables with focus in the muscle and joint stiffness value, it was found that some of the output variables could be used to approximate the stiffness value. These variables are the calculated muscle length (**plen**), muscle force (**fmus**), joint orientations that are in joint-specific Euler angles (**phing**), and the unit-moment vector. The moment acting on the joint is the result of multiplying the unit-moment vector with the muscle force. The unit-moment vector defines the direction of the moment taking into account the muscle moment arm. Thus multiplying the magnitude of the muscle force with it would result in the joint moment.

From these output variables, one could simply plot using Matlab the muscle force against the muscle length 4.4 and the joint moment against the joint angle 4.5. Appendix 2 (page 75) provide examples of Matlab commands to read the output data matrices and to plot it. The muscle stiffness value may be the incremental (from one time step to the next one) muscle force divided by the incremental muscle length. The joint stiffness could be calculated by comparing the increment of joint moment with the joint orientation, whether it is from every single muscle element or the resultant (there might be more than one muscle driving the joint rotation).

This way of deriving muscle and joint stiffness is prone to error because: 1) The results depend on the assumed optimization criterion to estimate the muscle force given a system with redundant actuator (much more muscles than the number of joint). 2) It has been explained in Chapter 3 that even if the muscle dynamics is available, the stiffness value is intricate to derive. Because of that, this approach is abandoned. Instead, the best approximation of muscle stiffness, which induces the joint and end-point stiffness, is deriving it from the muscle dynamics equations used in the model and then include criteria affecting the stiffness value such as activation, the movement range, contraction history, etc.

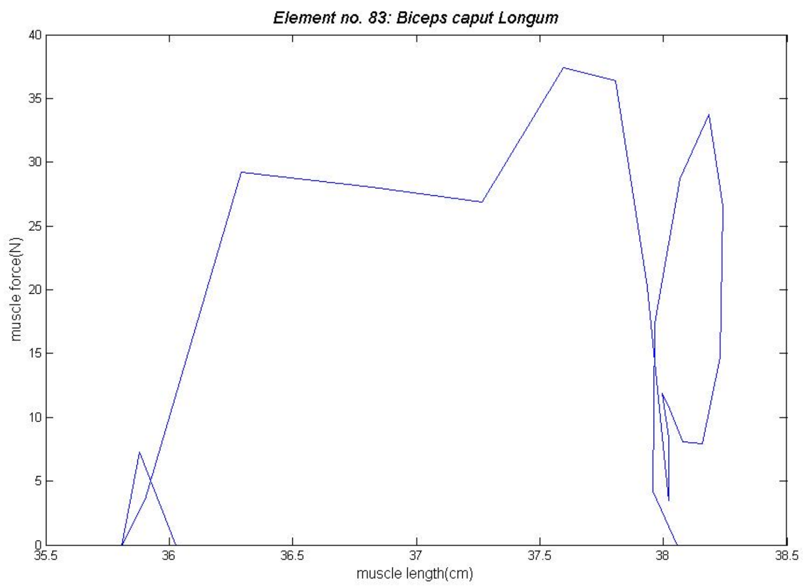


Figure 4.4: Force-length plot for muscle Biceps, caput longum

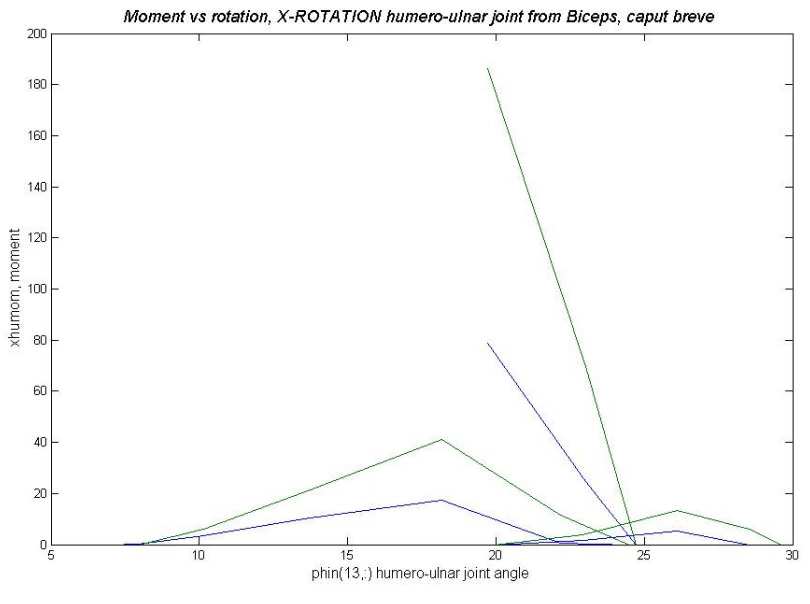


Figure 4.5: moment-angle of humero-ulnar joint plot for muscle Biceps, caput breve

4.2.2 Forward dynamics

In forward dynamics, one could find out the optimum and stable solution of motion given the neural input or CNS control 4.6. In order to do so, a muscle dynamics is included to transform the neural input into muscle and joint forces and eventually, the motion. The drawback is the expensive optimization (time consuming). The advantage is that there is no kinematic assumption.

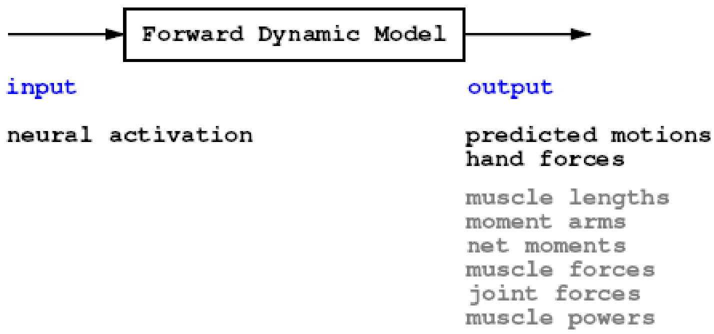


Figure 4.6: The forward dynamics of DSEM [8]

The forward dynamics had been attempted using DSEM but was put on aside because of the computation cost. In the DSEM program package received from TU Delft, only *inverse simulation* and also *IFDOC simulation* are available. IFDOC is an inverse simulation optimized using a forward dynamic simulation and with additional control block to ensure the neural-input is optimal. The next two sub chapters discuss this simulation.

4.2.3 Inverse-Forward dynamics optimization (IFDO)

The *Inverse-Forward dynamics optimization* (IFDO) is basically estimating the optimum neural inputs from the recorded-motion. As shown in figure 4.7, the estimated force from the inverse dynamics is put to optimizer to get the optimum force for the inverse muscle model. The inverse muscle model will calculate the corresponding neural inputs. The forward muscle model is a feedback block taking in the calculated neural input and calculating the Fmin and Fmax. Fmin and Fmax are calculated by integration of the muscle states to the next time-step, and ensuring *physiologically feasible bounds* for the optimization [8].

The schematic muscle dynamics that is Hill based muscle model is shown in figure 4.8. This is the basic of the muscle model block. The muscle dynamics used in DSEM is based on the work of Winters and Stark discussed earlier in section 3.2.1. The *inverse muscle dynamics* is also based on the work of Winters and Stark but derived by Happee in [16]. With this inverse muscle dynamics, one could calculate neural inputs from recorded motions. The muscle model equations derived by Happee are force-length based, instead of torque-angle based as presented in section 3.2.1. This force-length relationship is used in DSEM subroutine (**muclemod.f**).

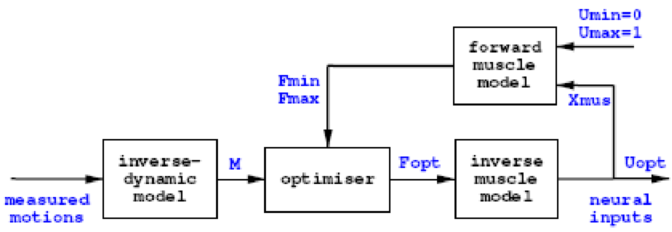


Figure 4.7: Inverse-forward dynamics [8]

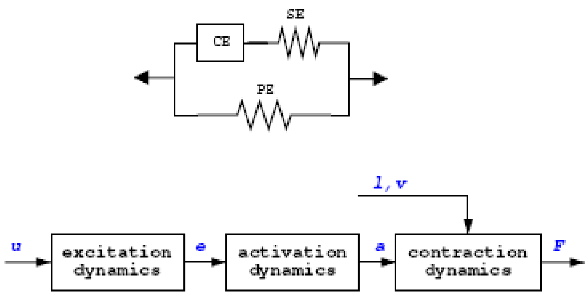


Figure 4.8: The muscle model (based on Hill's Model) for the muscle dynamics block in IFDO routine [8]

The differentiated force-length equations from the dynamics are presented in the next section because in fact, only IFDOC simulation in addition to the inverse simulation is in the DSEM package. It can be expected that the muscle is modeled modelled by a contractile element, a passive series element and a passive element just as explained in section 3.2.1.

4.2.4 Inverse-Forward dynamics with Controller (IF-DOC)

The concept of IFDO simulation has been introduced. Now, as the *forward muscle model* ensures physiologically feasible solutions for muscle force optimization, the added *controller* ensures calculated neural inputs reproduce the measured motions. It is shown in figure 4.9 where the controller block takes as an input the difference between the recorded motion input and the simulated motion. Then, it calculates the equivalent (ΔM) which is added to the output of the *inverse dynamic model*. In the DSEM IFDOC simulation routines, this is a subroutine that equalizes the forward and inverse muscle model (*equalize forward inverse* subroutine).

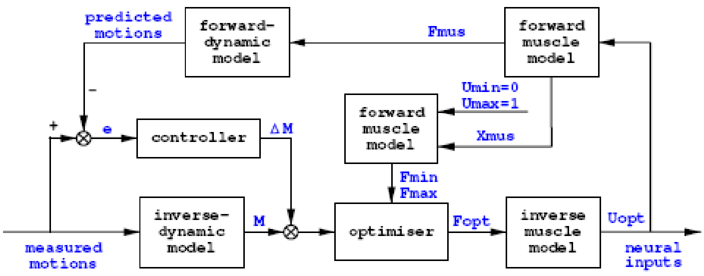


Figure 4.9: Inverse-forward dynamics with controller [8]

An application is the computer assisted surgery (CAS). In CAS, IFDOC can be run to obtain the neural inputs from healthy motion input 4.10. Then, using this optimum neural inputs, virtual surgery could be done with only forward dynamic simulation 4.10 without any optimization or fast simulations.

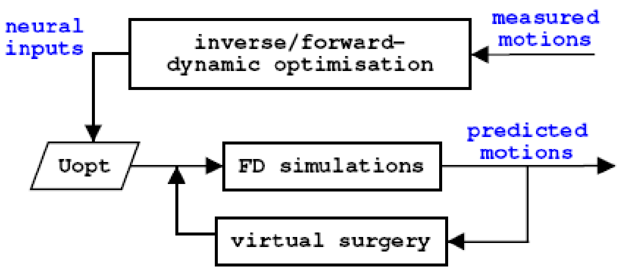


Figure 4.10: IFDOC simulation for Computer Assisted Surgery (CAS) [8]

In calculating the muscle stiffness, the equations in *musclemod* subroutine are presented here. *Musclemod* subroutine is evoked by the *forward muscle model* subroutine, to estimate the minimal and maximal muscle force. The *inverse muscle model* subroutine also calls *musclemod* in order to calculate the optimum neural input (u_{opt}). This process has been explained and also shown schematically in figure 4.9. In the *musclemod* subroutine, where those equations are written, the input variables are the muscle length and the excitation (exc)and the state variables are the CE length l_{ce} and the active state (act). The active state variable act is addressed with *Nea* by Winters and Stark (equation 3.4). The total muscle force is defined as the sum of several force components,

$$F_m = F_{max} \cdot (F_{se} + F_{pe} + F_{pv}) \tag{4.1}$$

In results the muscle stiffness, $k_m = \frac{d}{dl_m} F_m$, is

$$\frac{d}{dl_m} F_m = \frac{d}{dl_m} \left(F_{max} \cdot (F_{se} + F_{pe} + F_{pv}) \right) \tag{4.2}$$

Where,

F_m : The total muscle force

F_{max} : The maximum isometric force, which is proportional to the PCSA and not a function of muscle length in current DSEM.

F_{se} : Series element (SE) force. which is equal to contractile element (CE) force

F_{pe} : Parallel element (SE) force

F_{pv} : Passive viscous (PV) force, which is $F_{pv} = -PV \cdot V_m$

l_m : muscle length

V_m : muscle velocity

The followings are equations for force-length relationship for CE, SE, and PE derived by Hapee [16]. However, the followings are derived from the musclemod subroutine. Since $F_{ce} = F_{se}$, equation 4.5 is the stiffness for CE and SE.

$$F_{se} = \frac{1}{exp\,(SEsh) - 1} \cdot \left[exp \left(\frac{(l_{se}) \cdot SEsh}{SExm} \right) \right] \tag{4.3}$$

With $l_{se} = l_m - l_{ce} \cdot \cos(alpha) - l_t$ and using *the chain rule*, the following holds,

$$\frac{d}{dl_m}\left[exp\left(\frac{(l_{se}) \cdot SEsh}{SExm}\right)\right] = \left(\frac{SEsh}{SExm}\right) \cdot \left[exp\left(\frac{(l_{se}) \cdot SEsh}{SExm}\right)\right] \tag{4.4}$$

Finally,

$$\frac{dF_{se}}{dl_m} = \frac{1}{exp\,(SEsh) - 1}\left[\left(\frac{SEsh}{SExm}\right) exp\left(\frac{(l_{se}) \cdot SEsh}{SExm}\right)\right] \tag{4.5}$$

The derivation is the same for PE but with shape parameter and maximum displacement addressed as PEsh and PExm.

$$\frac{dF_{pe}}{dl_m} = \frac{1}{exp\,(PEsh) - 1}\left[\left(\frac{PEsh}{PExm}\right) exp\left(\frac{(l_m - l_{p0}) \cdot PEsh}{PExm}\right)\right] \tag{4.6}$$

For the passive viscous force, $F_{pv} = -PV \cdot V_m$ with PV is a constant and V_m is the muscle velocity. With the present understanding of V_m, its differentiation is kept as $\frac{d}{dl_m}F_{pv} = -PV \cdot \frac{d}{dl_m}V_m$.

At first glance, these equations do not seem to have the equation of Winters and Stark for active contractile element (equations A.1 and A.1 in Appendix 1, page 73). In the *musclemod* subroutine, the equation of lengh-tension relation (*ltrat*) is the equivalent of torque-angle relation (MXrat (q)) of Winters and Stark. The length-tension relation (*ltrat*) is used to calculate SE force F_{se} if the contraction is neither eccentric nor concentric. It is worth mentioning that there are modifications observed in *musclemod* subroutine of DSEM from the equations of Happee in [16]. This modification is the inclusion of pennation angle (*alpha*) in calculating *ltrat*, which is $ltrat = \cos(alpha) \cdot exp\left(-\left(\frac{l_{ce}-l_{ce0}}{l_{cesh}}\right)^2\right)$.

In *musclemod* subroutine, when the contraction is neither eccentric nor concentric, the following holds for SE force.

$$F_{se} = act \cdot ltrat \cdot MVml \tag{4.7}$$

Or,

$$F_{ce} = F_{se} = act \cdot \cos(alpha) \cdot exp\left(-\left(\frac{l_{ce} - l_{ce0}}{l_{cesh}}\right)^2\right) \cdot MVml \tag{4.8}$$

From here we can find the slope of the active force-length plot $\frac{d}{dl_{ce}}F_{ce}$,

$$\frac{d}{dl_{ce}}F_{ce} = act \cdot \cos(alpha) \cdot \frac{d}{dl_{ce}}\left[exp\left(-\left(\frac{l_{ce}-l_{ce0}}{l_{cesh}}\right)^2\right)\right] \cdot MVml \quad (4.9)$$

Then, just like equation 3.3, the stiffness of CE is

$$k_{ce} = act \cdot \cos(alpha) \cdot \left(-\frac{2l_{ce}-2l_{ce0}}{l_{cesh}^2}exp\left(-\left(\frac{l_{ce}-l_{ce0}}{l_{cesh}}\right)^2\right)\right) \cdot MVml \quad (4.10)$$

With,

act : activation or active state

$MVml$: max. eccentric force

l_{cesh} : shape length-tension relation

l_{ceO} : optimum length of CE (the difference between initial length and the tendon length)

Unfortunately, when running the IFDOC simulation using the available input, which is given as a sample in the DSEM program package, there is an error message before the end of the simulation. This problem could not be solved due to lack of time of DSEM experts at TU Delft. In results, the muscle stiffness calculation, i.e. the differentiation of muscle force with respect to muscle length presented here, is not yet implemented in the program. Another reason is that the program is already complex and there are still some unknown parameters and calculations, which are difficult to see just by reading the subroutines and trying to track down the meaning of those unknowns.

Finally, because the concept of calculating the muscle stiffness, especially in point of view of the muscle dynamics used in DSEM, has been studied, the application in the program should be straight forward with the help of the people knowing the program in details. In the near future, working with the DSEM experts would help in applying the calculation inside the programs faster and without the risks of clashing with the already existing variables and subroutines.

Chapter 5

Conclusion

Toward the incorporation of human arm stiffness calculation in the biomechanical model DSEM, the theoretical fundamentals have been laid out in this thesis. These fundamentals are namely the mechanical properties of the skeletal muscles, the mathematical mappings between the three components of the arm stiffness i.e. the muscle, joint and end-point stiffness, and the basics of dynamics simulations of DSEM i.e. inverse dynamics and inverse-forward dynamic optimization with control (IFDOC).

Dealing with skeletal muscles as actuators of the system and trying to apply it analogously in a robotic system that is DLR HASy, the mechanical properties of skeletal muscles were reviewed and discussed. As an actuator, a skeletal muscle is non-linear in most of its ROM. As a material,a skeletal muscle is anisotropic; thus, the exerted muscle force and displacement should take this anisotropy into consideration. The arm, as a musculoskeletal system is a complex kinematic chain and thus, it is not easy to quantify its mechanical properties.

From the observed mechanical properties, the stiffness value could be derived. It is important to note that when the effect of inertia and damping is taken into account, the term *mechanical impedance* should be used. The role of the muscle stiffness in postural stability has been discussed in the thesis. The study of stability has risen another mechanism to produce the muscle stiffness that is the reflex feedback gain. The other mechanism is the effect of intrinsic parameter i.e. the intrinsic viscoelasticity and voluntary activation. The effects of activation level to the muscle stiffness value have also been discussed. The presence of activation actually lifts up the level of complexity. In the real human motion, even a simple motion might include complex activation-deactivation dynamics and neuromusculoskeletal control system.

DSEM as a biomechanical model is currently able to do inverse dynamics simulation. The optimized inverse dynamics simulation including muscle dynamics (IFDOC simulation) is also available but has an error message before the end of simulation. Theoretically, deriving the stiffness value from the muscle dynamics inside the IFDOC simulation of DSEM allows us to examine the instantaneous stiffness. Also, the optimized neural input is calculated using the inverse muscle model in IFDOC. Using the inverse dynamics simulation, one could only estimate the stiffness from the output data i.e. muscle length, muscle force, joint unit-moment vector, and joint angle. These values are calculated and stored in matrices with the columns being the time step. Finally, the forward dynamics is unfortunately not available in the current DSEM program package.

It is required by HASY that the dynamics and non-linear joint stiffness should be calculated given a muscle activation information (forward dynamics) or a predefined motion (inverse dynamics). The IFDOC simulation of DSEM can be used by HASy for its inverse dynamics tasks. IFDOC simulation can be used to derive reliable the muscle and joint stiffness values because it includes the muscle dynamics, an optimization ensuring the physiological feasibility, and a control process ensuring the simulated motion is the same as the recorded motion input. The calculated neural activation would still be interesting for studying the human arm better, which is one of the aimed application of HASy. For example, given a recorded motion as a goal e.g. balls catching, object manipulation, the IFDOC simulation could be used to derive the muscle and joint stiffness and also the neural activation. The calculated joint stiffness is used as an input to HASy and HASy should be able to do the same motion as the one recorded.

Future works

The direct continuation after this thesis would be to include in the program the muscle stiffness calculation based on the muscle dynamics used in DSEM. In order to do so, a meeting with Prof. Van der Helm in Delft is planned to better understand the program and apply the stiffness calculation in the program. After the muscle stiffness calculation is incorporated, further works would be applying the transformation from muscle to joint stiffness and end-point stiffness in DSEM subroutine. The simulated joint stiffness, calculated by DSEM, might need to be transformed into HASY's joint stiffness. It is expected that even though HASy joints are anthropomorhic, the coordinates definition might be different from what is defined in DSEM. Until now, the exact kinematic definition of HASy is not yet available.

Also, it is important to highlight a few possible way to prove the integrated DSEM-HASy system. First is to compare the motion created by HASy, the motion of the human, and the motion simulated by DSEM. It will be interesting to compare the observed HASy's end-point stiffness with the end-point stiffness simulated by DSEM. This is another reason why the end-point stiffness should also be incorporated in DSEM in addition to the muscle and joint stiffness. One could also compare the predicted optimum neural input from IFDOC simulation with the recorded EMG (Electromyography) data taken during the motion recording.

Applications

The application of DSEM in developing HASy could also improve other technology. The comparison between the calculated neural input from IFDOC simulation and the EMG recording will be beneficial to both biomechanical modelling and bioimaging technology that is to improve both the IFDOC simulation program and the EMG technology. This would also lead in more understanding of the neuromusculoskeletal system.The same could be done by comparing the effect of limb stiffness in the stability of the simulated limb with the effect in the stability of the real limb and perhaps also HASy.

It has been mentioned, an anthropomorphic system would give a wider horizon of human-robot interaction and also robot flexibility. The space robotics which requires such anthropomorphic manipulation, safe astronaut-robot interaction would benefit from the development of an anthropomorphic system.

This study is also relevant to the space health and space medicine topics. In the human space flight field, there is a high interest in studying the muscle behavior in microgravity. There is an alteration of muscle activity and also stiffness observed in microgravity. In addition, muscle atropy[1] and degeneration are also observed. The study of skeletal muscle stiffness can be useful for the design and development of space sports, improvement of austronaut's health, and development of the *integrated* and *closed life support system*. Personally, it will be interesting to study the muscle stiffness using also a biomechanical model but targeting the microgravity scenario. This should bring a two-way advance. For example, the space medicine discipline can make use the results for designing the space sports for the astronauts and also to support other activities such as EVA [2], ergonomic design for the working-space in microgravity environment, etc. The application in space medicine may also help the biomechanists to understand better the neuromusculoskeletal system.

[1] Reduction of strength
[2] Extra Vehicular Activity

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Appendix 1: Muscle model equations from Winters and Stark (1985)

The contents of this Appendix are from [39]. First, the gauss-equation for active element (CE) is the following:

$$MXrat\,(q) = exp\left[-\left(\frac{q - MXoo}{MXsh}\right)^2\right] + MXsl \cdot q \tag{A.1}$$

$$Mh\,(q) = Nea \cdot MXrat\,(q) \cdot Mmax \tag{A.2}$$

The torque-angle equation for the passive element (PE) is:

$$M_{pe}\,(q) = ko \cdot q + bo \cdot v + k1(exp(k2 \cot q) - 1) \tag{A.3}$$

For the series element (SE), the equation is just the equation A.3 without the linear viscosity element, *bo*, as the viscous term is approx.0.1 Nms/rad.

$$M_{se}\,(q) = ko \cdot q + k1(exp(k2 \cdot q) - 1) \tag{A.4}$$

The constants k1 and k2 for parallel element PE are:

$$k1 = \frac{Mmax}{exp(PEsh)}$$

$$k1 = \frac{PEsh}{PExm}$$

And the same for SE but with SEsh and SExm instead of PEsh and PExm.

Parameters description:

Nea : Neural activation

Mmax : maximum moment

PExm : displacement at Mmax

PEsh : parallel element shape parameter, with a higher value resulting in higher curve concavity

MXoo : angle at maximum moment

MXsh : Gaussian-type "shape" function

MXsl : "slope" parameter, only sometimes needed

Maximum Extension (SExm) for SE:

SExm(fast) = 7 %

SExm(slow) = 6 %

SExm(tendon) = 4 %

Series Elastic Shape parameter (SEsh):

SEsh(muscl) = 1.5 + 3.0 * (slow muscle fibers fraction)

SEsh(tendon) = 1.5.

Appendix 2: Plotting the output of inverse dynamic simulation with Matlab

After running an inverse dynamic simulation of abduction motion one could give a Matlab command:

```
>>plot(plen(83,:), fmus(83,:)
```

One could plot the force exerted by the muscle element number 83, muscle Biceps, caput longum against its length (Figure 4.4). Table 4.1 shows the element numbers belong to the 31 arm muscles in DSEM. In fact, one could do the same for all 30 muscles (139 elements). Doing the same for muscle Tapezius, scapula part (Element number 1-11), there should be plots from the eleven representative elements.

In this simulation there are 26 iterations (column), the maximum number of muscle elements is 200 in rows (only 139 rows have values in it) and that mxhu is a matrix of the unit-moment vector of humero-ulnar joint and fmus is muscle force matrix, both with dimension 200 x 26, the following calculates the joint moment.

```
>> for i=1:26;
 >>  for j=i:200;
 >>       % humero-ulnar moment, only in x direction
 >>       xhumom(j,i)=mxhu(j,i)*fmus(j,i);
>>    end
>> end
```

After finding the moment of the respective joint, one could observe the joint orientation matrix, **phin**. For the humero-ulnar joint, the 13^{th} row of phin(21,26) is the orientation angle, calculated for each simulation step. When one gives the command:

```
>>plot(phin(13,:), xhumom(84,:), phin(13,:), xhumom(85,:))
```

The plot will be as shown in figure 4.5. The plot shows the moment created by the two elements representing muscle Biceps, caput breve.

The following can be written to calculate the moment in the SC-joint for the same simulation with allowable muscle number is maximum 200 and there are 26 iterations. It can be observed that for SC-joint, the ball socket joint, there are three components defining its moment (x,y,z). These are not in cartesian system but in Euler angle representation. In fact, the orientation of SC-joint is defined as y-z-x Euler angles, measuring the Scapula with respect to Clavicula. The orientation matrix is therefore, phin(4,:) for y-rotation, phin(5,:) for z-rotation, and phin(6,:) for x-rotation. Then, one may plot the moment in x-direction with respect to the x-rotation angle and the same can be done for the other orientations.

```
>>for i=1:26;
  >>  for j=i:200;
    >>    % SC-joint moment, only x,y,z directions
    >>    xscmom(j,i)=mxsc(j,i)*fmus(j,i);
    >>    yscmom(j,i)=mysc(j,i)*fmus(j,i);
    >>    zscmom(j,i)=mzsc(j,i)*fmus(j,i);
  >>  end
>>end
```

In addition to humero-ulnar and SC joint, there are another four joints, of which moments and orientations can be plotted. They are the AC, GH, Ulnar-radial, and Radio-carpal joints.