A Joint-level Model of Fatigue for the Postural Control of Virtual Humans

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Abstract

In this paper a model is proposed to evaluate individually the normalized *fatigue* at the joint level. More specifically our model distinguishes the two *antagonist muscle groups* acting for each *degree of freedom* (in short *dof*). The fatigue model parameters are normalized torques, joint strength and the maximum holding time that a posture can be maintained. Fatigue evolution is predicted taking into account how these parameters evolve over time. Fatigue levels can be exploited by a posture control algorithm.

Keywords: human body simulation, posture control, fatigue, computer animation.

1 Introduction

Achieving a task through the postural control of the human body is very difficult since the musculo-skeletal structure has so many interrelated components involved in the movement production. In addition, complex factors as posture-dependent muscular strength, fatigue and state of mind, influence the posture choice and temporal evolution among a generally infinite solution space.

Computer animation systems that work specifically with humans should produce realistic animation in all the sense of the word, including fatigue sensation when doing tasks as lifting or pulling some external load.

One obstacle for fatigue assessment of human body's joints is the diversity of data coming from different studies relative to muscular strength of human body. We have used those coming from Chaffin studies that are gathered from several researchers [Cha88]. An additional problem is the specific scope of these studies, sometimes restricted to specific tasks (i.e lifting). The purpose of our model is to define the set of parameters and relations between them that define fatigue at joint level. As we present in the following sections these parameters are joint strength, current torque and maximum holding time that a posture is sustained until the state of exhaustion where maximum fatigue level is reached.

The organization of the paper is as follows. In section two we deal with the related work. In section three, we present fatigue at joint level versus fatigue at muscle level. In section four, the fatigue model is described. In section five, we present the 3D environment used to do the simulations. Lastly, concluding remarks and future work are given in section six.

2 Related work

A study from Hashimoto defined fatigue from different points of views. Physiologists consider fatigue simply as a decrease in physical performance. Psychologists consider it as a condition affecting the mental process. Ergonomists and physicians lay stress on consequences of fatigue [Has71]. For us, the concept of fatigue lies half way between the physiologists and physicians viewpoints.

In a fatigued muscle the intracellular pH level decreases. A recovery phase is necessary for the pH level to increase again. Giat modeled the fatigue of quadriceps muscle obtaining the relationship between the intracellular pH and force exerted by the muscle. The force-pH relationship was fitted by an exponential function [Gia96].

In the field of human postural control, [Lee93][Lee93] and Badler [Bad93] implicitly searched for the postures least prone to local joint fatigue by ensuring that the ratio of joint torque to joint strength is minimal. However the explicit modeling of fatigue evolution over time is not treated. More recently Komura [Kom99] [Kom00] used Giat's model to simulate the fatigue of each individual muscle for a Hill-type musculotendon model of the leg proposed by Delp [Del90] . The study from [Kom99] establishes the dynamic equation relating a muscle exertable force to its current length, contraction velocity, activation level and normalized fatigue. The normalized fatigue term is computed differently depending on the current state of the muscle, either *fatiguing* or *recovering*. The authors propose to switch between the two states based on the activation level: over a 50% activation level the muscle is fatiguing while under it is recovering. This model has been used for the motion retargeting of the lower body in [Kom00].

Muscular strength evaluation has also been used to determine the relationship between degraded muscular strength and alteration of lifting motion strategy [Zha02]. It can be used to assess postural deficiencies in daily activities to reduce the number of industrial workers suffering from back injuries. Another complementary research topic tries to determine how a general state of mental fatigue decreases performances like attention and response time [Jon98].

Our work is restricted to the evaluation of the muscular fatigue at the level of muscle groups as defined in the next section. We target this coarser level - compared to Komura's studies - to allow its integration into a full body posture control system running at interactive rates [Bae02].

3 Fatigue at joint level versus fatigue at muscle level.

We distinguish two points of view with respect to fatigue assessment: fatigue at joint level or at muscle level.

Fatigue at joint level evaluates the fatigue produced by a *group of muscles* acting on a joint. External forces acting on a body segment under gravity produce load moments at body joints. These load moments can be compared to muscle strength moments and thus provide a means to evaluate along a simulation how stressed the joint is. Strength equations provide us the muscle-produced moment strength at each joint. Because muscles often span two (or more) joints, the angle at adjacent joints must be considered in predicting some strength values. In a joint as complex as the shoulder, the motion in the sagital plane is produced by a group of muscles that also contribute to other joint movements (e.g. the elbow).

Assessing fatigue at muscle level is a tremendous task as the human body contains more than 650 individual muscles. The movement of the sole shoulder joint is produced by fifteen muscles and as mentioned before, a single muscle can participate in some movements as the prime mover and in other as assistant muscle. For these reasons, our model is defined at the joint level as described now.

4 Fatigue model

We propose to calculate fatigue at joint level, more precisely at *muscle groups* level. To do that, we need the concept of *half-joint pair*. We split each single *dof* joint (*dof* is short for *degree of freedom*) into two coordinated *half-joints* constituting a *half-joint pair*. The two half-joints of a pair simulate the activity of the group of muscles associated to <u>one</u> degree of freedom but acting in opposite *pulling* directions. They are said to be *antagonist*. This distinction allows us to calculate and visualize independent fatigue variables for each antagonist muscle group. It is also exploited within our IK architecture [Bae02].

For example, elbow flexion and extension movements are produced by two antagonist group of muscles. That's the reason why there are two strength curves for elbow in the sagital plane [Cha88]. In Figure 1 the elbow joint - with its constant range of motion [θ_{E_min} , θ_{E_max}] - is split in two half-joints with dynamic limits as follow (θ_{E_cur} represents the elbow current state):

- half-joint θ_{E1} works in the same direction as the original joint (flexion); it has a range of motion [θ_{E_cur} , θ_{E_max}]
- half-joint θ_{E2} works in the *opposite* direction as the original joint (extension), thus its range of motion is [θ_{E_cur} , θ_{E_min}].

The second half-joint is implemented as a transformation concatenated *after* the first half-joint's transformation, thus making its range of motion relative to the first half-joint's state with the following limits: [0 , θ_{E_cur} - θ_{E_min}]. Figure 1 illustrates this design choice.

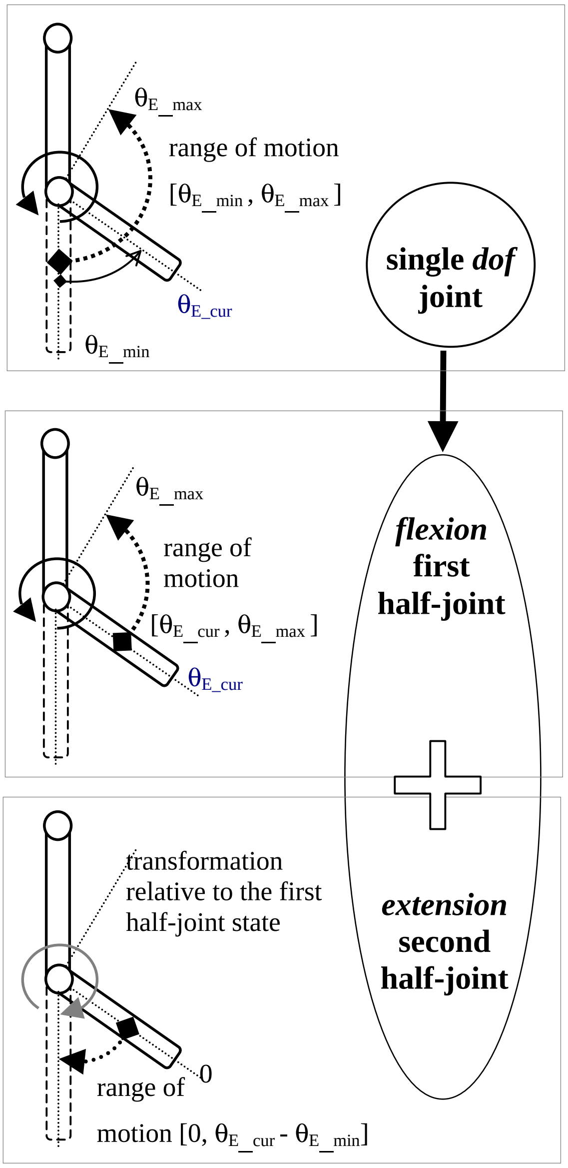

Figure 1 A single dof joint (the elbow) is split in two half-joints (constituting a half-joint pair)

Figure 2 gathers the parameters of the proposed fatigue model. The joint strength and the current joint torque are used to calculate a value of normalized torque. The normalized torque is used to calculate the maximum holding time that the posture can be sustained (in a static context). Finally the current value of the maximum holding time and the time increment are considered to update the joint fatigue. This process operates while trying to achieve user-defined goals with our IK solver [Bae02].

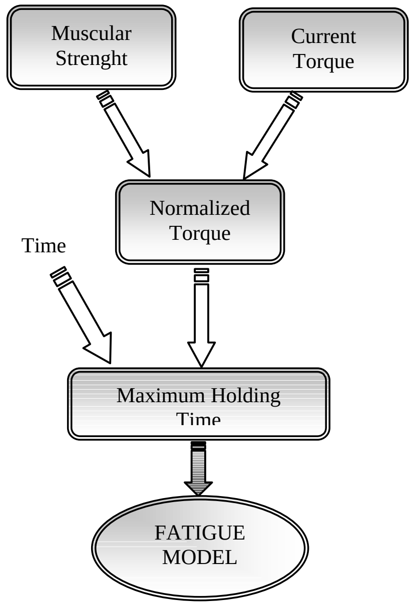

Figure 2 Fatigue model parameters at the half-joint level (i.e. muscle group).

We now explain how each parameter is computed.

Muscular strength

A *loading moment* is the moment, with respect to a specific motion axis, caused by the body segments mass and by external forces. This moment is counteracted by a moment of equal magnitude but opposite direction exerted by the active muscles. Loading moments can be used to describe the load over time, to compare different ways of performing a task or to study the effects of various means of load reduction.

Muscular strength is the maximum capacity that a group of muscle has to produce force. The simplest way to relate loading moment to maximum capacity (muscular strength) is to normalize it by the latter. Most studies express their findings in these normalized quantities.

The main function of a muscle is to produce a force along the muscle line of action. Depending on the current posture which influences the muscle lever arm, this force causes a moment with respect to a joint axis. This explains that, for a group of muscles acting in a given direction (e.g. flexion or extension), the maximal exertable torque is not constant but depends on the current posture [Kul85].

An important aspect to take into account in a fatigue model is the decrease of muscular strength as a function of fatigue as the transmission of muscle activation is inhibited at the cellular level [Tsa69]. Clarke has shown that the strength decreases by an amount of 30% when fatigue raises from a fatigue level of zero to the maximum fatigue level [Cla66]. The strength starts decreasing when the fatigue reaches a certain threshold. Available data make us estimate this threshold level to be around 40% of the maximum fatigue level (exhaustion).

Current Joint torque

We work in the static equilibrium hypothesis, so the joint torques can be computed from the Jacobian transpose using the principle of virtual work [Cra86]. The Jacobian transpose maps the cartesian force active at an effector into an equivalent joint torques:

$$\tau = \sum_i J_i^T F_i + J_G^T P$$

where F represents an external Cartesian forces, P represents the body weight acting at the center of mass, J_i^T the transposed Jacobian for the end effector i, J_G^T is the transposed Jacobian dedicated to the position control of the center of mass [Bou96] and τ is a vector storing joints torques.

Then, for each joint *i*, the normalized torque $T_{N\,i}$ is the quotient of joint torque τ_i and joint strength st_i :

$$T_{N_i} = \frac{\tau_i}{st_i}$$

Maximum holding time

Manenica stated that there was a relationship between the force a muscle exerts and the maximum holding time that the posture can be maintained [Man86]. The maximum holding time is a function of the normalized torque:

$$mht = f(T_N)$$

An experimental study of several groups of male and female highlighted that there were not so many differences between force-time relationships in different muscle groups. Therefore a general force-time relationship expressed as a regression line, valid for several muscle groups, was obtained:

$$mht_i = \exp(2.70 - 0.0448\ T_{N_i}) \qquad (1)$$

where mht_i represents the maximum holding time that joint *i* is able to maintain the posture and $T_{N\,i}$ represents the normalized torque. Time is expressed in minutes.

By definition the fatigue level is expressed as the holding time *ht* normalized by maximum holding time *mht* :

$$fatigue\_level = \frac{ht}{mht} \qquad (2)$$

We consider a more general case than the biomechanics studies as we allow the posture to change slowly over time. In that context we still want to estimate how the fatigue evolves over the elapsed time. For that purpose we propose a variational expression of fatigue where the fatigue variation is a function of the time increment duration Δt and of the current maximum holding time *mht* :

$$\frac{ht_k}{mht} = \frac{ht_{k-1}}{mht} + \frac{\Delta t}{mht} \qquad (3)$$

Formula (3) defines the fatigue level at time k as fatigue level at time k-1 plus a term that expresses an additional fatigue increment produced during Δt. In such a way we extend the scope of formula (2) due to the possible variation of *mht* over time depending on the context (calculated with (1)). Several studies have concluded that fatigue appears when the normalized torque is above a threshold of 15% of the muscular strength [Roh60]. For this reason we propose to scale the fatigue increment with a factor representing the gradual manifestation of fatigue. This is the fatigue factor (F_f) as shown in the following equation:

$$fatigue\_level_k = fatigue\_level_{k-1} + \frac{\Delta t}{mht} F_F \qquad (4)$$

F_F is defined as follows:

$$\begin{aligned} 0. \le T_N \le 0.15 &: \quad F_F = 0 \\ 0.15 \le T_N \le 0.30 &: \quad F_F = 6.6 T_N \\ 0.30 \le T_N \le 1.0 &: \quad F_F = 1 \end{aligned}$$

Equation (4) is now extended with a negative term, called the *recovery decrement*, approximating the effect of a static recovery produced by a period of rest. Milner carried out an investigation to observe the relationship between holding time, rest and recovery [Mil86]. He has shown that, for almost all fatigue levels except exhaustion, the minimum duration of recovery *mdr* was equal to the current fatigue level multiplied by the maximum holding time. Therefore, we exploit formula (1) to evaluate *mdr* using a floating average of the torque to reflect its recent history. The window size of the floating average depends on the current instantaneous torque. Basically, if the instantaneous torque is very low, we use a wide window to compute the average torque. For higher instantaneous torques the window size decreases to be more sensitive to the recent past. Finally for even higher values the recovery term is filtered out by a recovery factor R_F that is function of normalized torque(T_N). The idea is that recovery can start for small values of the normalized torque, so R_F is defined as follows:

$$0.15 >= T_N >= 0 : \qquad R_F = 1 - 6.6T_N$$

$$1.0 >= T_N > 0.15 : \qquad R_F = 0$$

Finally, the formulation of fatigue level is given by:

$$fatigue\_level_k = fatigue\_level_{k-1} + \frac{\Delta t}{mht} F_F - \frac{\Delta t}{mdr} R_F$$

5 Simulation Environment

This section shows the simulation environment used to assess the fatigue model in the arm case study. The arm works in the sagital plane along the flexion-extension mobility axis for the shoulder and the elbow joints (this makes a two dimensional joint space). Inverse Kinematics is used for postural control to achieve a simple reach task [Bae98][Bad93]. A one dimensional reach task allows us to evaluate an additional task of torque fatigue minimization. During the simulation, the posture converges towards a solution at a rate compatible with joint velocity. The instantaneous values of the fatigue model are exploited to update the half-joints fatigue level for each time step.

Each half-joint fatigue level is then visualized around its corresponding rotation axis by means of a dynamic semicircle (see Figure 3). A half circle corresponds to the exhaustion of the associated half-joint (when the fatigue level is equal to one). The black arrow represents an external load that the arm is lifting.

Different strategies are possible to exploit the knowledge of fatigue levels within the posture control algorithm. First, at the lowest task level, a torque optimization term can exploit the redundancy of the articulated structure to improve the posture comfort [Lee93]] or to reduce the fatigue level of the most fatigued half-joints (see [Bae02b]for the expression of a joint torque minimization for a set external forces including the weight). The resulting gradient vector is simply projected on the null space of the reach task Jacobian [Bae98]. The problem with such an approach is the low priority of the task: even if an half-joint is exhausted, it may not be able to modify the posture towards a rest posture for that half-joint. So the alternate strategy is to introduce inequality constraints like those used to enforce joint limits [Bad93][Bae02]. In our context the inequality is dynamically created to enforce the torque minimization when a half-joint fatigue is over a predefined threshold. This approach "pushes" the realization of the reach task in the posture space that reduces the fatigue.

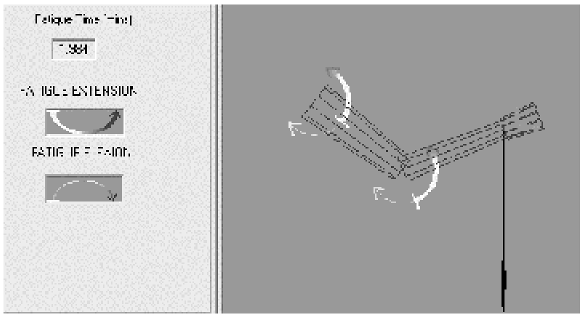

Figure 3 Fatigue visualization in the arm case study

6 Conclusion

In this work, a model for fatigue assessment at joint level has been presented. The model is embedded in a simulation environment which gives the user a feedback on fatigue levels by providing visual feedback around each rotation axis. The core of the system is a new approach for fatigue modeling based on studies in biomechanics. Contributions of the model are the reduced number of parameters used and the incremental consideration of time in our model. The lack of data from medical or biomechanical studies presently restricts us to deal with motions in the sagital plane. In particular these motions are lifting of external loads. As the present model considers the static case, a direction for future work includes the adaptation to the dynamic case.

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