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A passive movement method for parameter estimation of a musculo-skeletal arm model incorporating a modified hill muscle model☆

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ABSTRACT

In this paper we present an experimental method of parameterising the passive mechanical characteristics of the bicep and tricep muscles in vivo, by fitting the dynamics of a two muscle arm model incorporating anatomically meaningful and structurally identifiable modified Hill muscle models to measured elbow movements. Measurements of the passive flexion and extension of the elbow joint were obtained using 3D motion capture, from which the elbow angle trajectories were determined and used to obtain the spring constants and damping coefficients in the model through parameter estimation. Four healthy subjects were used in the experiments. Anatomical lengths and moment of inertia values of the subjects were determined by direct measurement and calculation. There was good reproducibility in the measured arm movement between trials, and similar joint angle trajectory characteristics were seen between subjects. Each subject had their own set of fitted parameter values determined and the results showed good agreement between measured and simulated data. The average fitted muscle parallel spring constant across all subjects was 143 N/m and the average fitted muscle parallel damping constant was 1.73 Ns/m. The passive movement method was proven to be successful, and can be applied to other joints in the human body, where muscles with similar actions are grouped together.

1. Introduction

In this paper we describe a passive method for parameterising the passive mechanical characteristics of human muscles in vivo. As an example, a study of the movement of the elbow joint and the procedure to obtain parameter values of an arm model incorporating the elbow flexor and extensor muscles as modified Hill muscle models is presented.

The focus for much biomechanical modelling has either been on body segment motion e.g. [1] or on the analysis of individual joint movements e.g. [2]. Whole body models used neural network (NN) or genetic algorithms (GA) to assign muscle forces and properties to individual muscle groups within the body from kinematic measurements e.g. [3,4]. However, limited anatomical and physiological data on individual joints and muscles were incorporated into these models. The majority of the modelling work on single joints has been aimed at

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understanding the motion around the joint and consequently the majority of the models generated were not predictive.

However predictive models are required for the design of prostheses or orthoses, in particular patient specific prostheses and orthoses. Orthoses and prostheses, including functional electrical stimulation (FES), are only one component of rehabilitation where achieving independence and performing activities of daily living (ADL) is the ultimate goal. Medically and therapeutically, clinicians often wish to use orthoses and prostheses to emphasise or de-emphasise parts of a current movement. Without detailed predictive models that are anatomically and physiologically meaningful, such changes to the movement cannot be incorporated into the prosthesis and orthoses and hence the overall strategy for the patient. Therefore, our goal was to generate musculo-skeletal models where components are physiologically and anatomically meaningful.

One of the stimuli for the current work is the development of models for the design of model based control system for FES systems. FES [5] has been used as part of rehabilitation strategies on spinal cord injured (SCI) patients for regaining movement functions, such as generating knee lock to allow standing, e.g. [6–9], or achieving balance by controlling ankle angle, e.g. [10,11]. Traditionally, many FES systems have used open loop on/off control, e.g. [8]. Such systems are simple to implement and do not require predictive models, however they were found to cause rapid muscle fatigue. For example, Chesler and Durfee's [12] study of maximum tension and fatigue under FES, showed that maximum tension reduced to 50% in about 15 s. Much of the work on closed loop FES controllers has been based on proportional integral derivative (PID), NN or GA controllers. In PID controllers [10], mechanistic models were used, but the bulk parameters had no anatomical and physiological meaning, and control systems were optimised to individual patients empirically. For NN, e.g. [3] and GA, e.g. [4], based controllers, machine learning techniques were required to obtain numerical values, but once again these had no anatomical or physiological meaning.

Irrespective of their purpose, biomechanical models usually contain unknown parameters, where values are determined through parameter estimation techniques. Traditionally, measurements from maximum voluntary contraction (MVC) have been used as part of the parameterisation of muscle models, e.g. [13–15], however, a problem arises if voluntary contraction is not possible, for example when working with SCI subjects. In these cases the MVC method cannot be used. As a solution, we propose an experimental method using passive movements, in which the muscles are completely relaxed and non-active, to obtain numerical values for the passive mechanical parameters in the muscle model. In the case of the bicep and tricep muscles, measurement of passive elbow flexion and extension was used for parameter estimation.

2. Background

Hill type muscle models [16], which are widely used in musculo-skeletal modelling, represent the muscle as a combination of mechanical components. Because these mechanical components model properties that result from a large number of microscopic events, which occur at the sarcomere level, it is not possible to measure the dynamic properties of these components directly for individual subjects in vivo. Therefore, the only approach to obtain parameter values is to use parameter estimation techniques, in which simulated data are fitted to measured data. Currently, few parameter values for the passive mechanical components have been published from studies where parameter values were obtained from measured data in vivo [2,17,18]. In one study [19], a promising approach to parameterising the classical Hill model was presented but these authors were unable to obtain parameter values, although the reasons for this are unclear. We have previously shown that the classical Hill muscle model is not structurally identifiable and therefore parameter values cannot be uniquely obtained through measurement [20]. As part of the same study, we showed that a commonly used modified version of the Hill muscle model [20–24] where there are no serial combinations within the parallel components was structurally identifiable if the internal component lengths of the muscle are known. These latter studies highlighted a further problem in that even where modified Hill muscle models had the same structure, there were inconsistencies between studies in the anatomical definitions of the model components. The anatomical definitions of the modified Hill muscle model used in this study follow those we outlined in the structural identifiability analysis [20] and are described in detail in Sections 3.2 and 4.1.

Our goal was to parameterise individual muscles or group of muscles that are similar in both action and geometry for a particular limb movement, in subjects who had no voluntary control of the muscles for that movement. Since the model follows the anatomy, with the bicep muscle group and tricep muscle working in opposing directions, separate experiments involving movement in each direction are necessary to parameterise the two muscles models. Venture et al. [21,25] have previously reported a similar passive technique for parameterising the elbow joint, however in their study only one experimental protocol was used and their final arm model became a simple 2nd order spring damper model that lacked an explicit muscle model.

The experiments in this study measured the action of passive elbow extension and flexion. Our preliminary results [26] showed the initial elbow extension experiment (denoted experiment 1 in this paper and described in Section 4.2) did not adequately describe the trajectory predicted by the model when maximum elbow extension was reached after 90° of movement. This only gave 0.6 s of data for parameter estimation, therefore in this paper, work on a modified version of the extension experiment (experiment 3) is described, in which a different upper arm orientation is used (see Section 4.4), providing a larger range of elbow angle movements (135°) for parameter estimation.

3. Materials

3.1. Musculo-skeletal model of the human arm

The two segment model shown in Fig. 1 is a representation of the human arm [20,26]. It has one degree of freedom around