A large-scale musculoskeletal model of the shoulder and elbow: The Delft Shoulder (and elbow) Model.

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INTRODUCTION

The shoulder is probably one of the most complex joint mechanisms of the human body. The shoulder consists of four bony elements: Thorax, clavicle, scapula and humerus. The connection thorax-clavicle-scapula is a closed chain. Due to the closed chain, the rotational and translational degrees of freedom of the scapula and clavicle are constrained, meaning that not every orientation can be achieved. In addition, the moments around the sternoclavicular (SC) and acromioclavicular (AC) joint are coupled, i.e. if a muscle exerts a moment around one of these joints, the scapulothoracic reaction force will change and the altered moment of the scapulothoracic reaction force around the other joint must be compensated by a second muscle.

In total 20 muscles and muscle parts are crossing the shoulder joints (SC, AC and Glenohumeral (GH) joint). Additionally, 11 muscles are crossing the elbow joint (flexion-extension in the humero-ulnar (HU) joint and pro-supination around the ulno-radial (UR) joint). Many muscles are bi- or tri-articular, and have large attachment sites. Parts of the larger muscles can contract independently from each other, as shown by EMG.

The goal of this presentation is to discuss a number of general properties that are important for a musculoskeletal model: Inverse vs. forward dynamic simulation, motion constraints, generic model vs. individual model, importance of length-tension properties, optimization algorithm and validation.

METHODS

A large-scale model of the shoulder and elbow has been developed (the 'Delft Shoulder Model', DSM), in which the 31 muscle and muscle parts are represented by 115 muscle lines of action. If appropriate, the muscle lines of action are wrapped over underlying bony contours, modeled as a sphere, cylinder or ellipsoid. The SC, AC and GH joint are each represented as a spherical joint with 3 rotational DOF. The scapulothoracic gliding plane (STGP) is modeled with the thorax as an ellipsoid, on which two points of the medial border are sliding. Hence, two DOF are constrained by the STGP. The reaction forces at the STGP are perpendicular to the surface of the ellipsoid. An additional DOF is constrained by the assumption that the conoid ligament is rigid. In general, inverse dynamic simulations are not appropriate to study the role of ligaments. Small errors in the recorded motions are amplified in the strains of the ligaments, and even more amplified by the steep stress-strain curve resulting in large errors in the passive forces. These passive forces must be generated by active structures like muscles. Hence, erroneous estimates of the muscle forces might result. The elbow joint is represented by one rotational DOF for the HU-joint, and one for the UR-joint. Hence, 8 dynamic DOF are modeled with the complete set of muscles crossing the joint. In addition, 9 kinematic DOF are defined (6 DOF for the thorax motion and 3 DOF for the wrist motion). The model is shown in Fig. 1.

A model is as good as the quality of the parameters. Geometric, inertia and muscle parameters are obtained in an extensive cadaver study of 14 shoulders of 7 specimen. Later, 5 elbows were measured. Effectively, three complete sets are used for simulations and comparisons. Recently, another specimen has been measured, in which also pennation angle, tendon length and optimum fiber length through laser diffraction was assessed.

The model is programmed in SPACAR, a multi-body dynamics program. Each morphological structure is represented by an appropriate element, of which the mechanical properties are known. By connecting all the elements, the mechanical properties of the whole mechanism can be calculated. The advantage of SPACAR is that it is relatively easy to change morphological parameters, or use a complete different cadaver set. Closed-chain mechanism do not pose any problem to SPACAR since the number of motion equations is reduced to the number of DOF, comparable with Kane's method.

The shoulder and elbow model is used inverse dynamically: Motions of the bony elements (including clavicle and scapula) are being recorded using an electromagnetic device (Flock-of-Birds) or camera system. Scapular and clavicular motions can only be measured statically. However, these positions can be extrapolated to the dynamic situation using regression equations of the scapulohumeral rhythm.

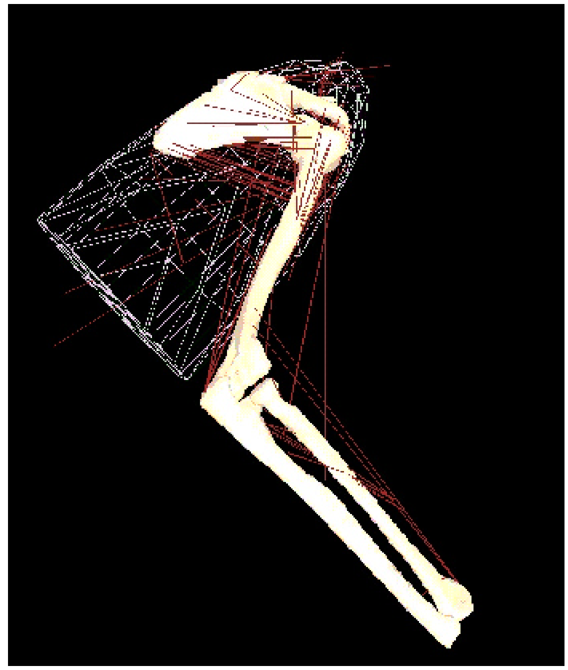

Fig. 1: *Visualization in SIMM of the DSM.*

In human motion control, proprioceptive reflexes originating from the muscle spindles and Golgi Tendon Organs (GTOs) are very important. In comparison, almost all robots are feedback controlled, but hardly any musculoskeletal model incorporates feedback. In an experimental application only including the shoulder, the length, velocity and force feedback pathways including time-delays have been implemented, representing the muscle spindles and GTOs. Time-delays have been represented by a second-order Padé approximation, which is sufficient to incorporate stability effects. The complete musculoskeletal model has been linearized, and a linear state space model resulted. A total number of 487 states were included in the system matrix. In the force feedback loop only the muscle dynamics and time-delay is present. Hence, the force feedback gains can be optimized for each muscle individually. Effectively, force feedback increases the bandwidth of the muscle dynamics, thereby reducing the intrinsic muscle visco-elasticity. A fixed relation between length and velocity feedback was assumed (10:1), which was shown to be near optimal in many other smaller optimization studies. A further reduction in unknown parameters was achieved by parametrizing the remaining length feedback gains by two parameters per muscle, resulting in 40 unknown parameters. Maximizing the endpoint impedance at the hand optimized these parameters.

RESULTS

Current applications of the model include clinical applications (diagnosis and Computer Assisted Surgery) and ergonomic applications. Clinical applications are the functional analysis of a glenohumeral endoprosthesis, tendon transfers, glenohumeral arthrodesis, partial muscle paralysis of spinal cord lesions in the cervical region, scapula fractures, clavicle fractures, etc. Ergonomic applications are the assessment of the load in the glenohumeral joint, for workers (garbage collectors, brick layers) and for wheelchair users. In addition, it has been analyzed which muscles are involved in energy production during the wheelchair propulsion motion.

Results are presented in a Matlab window, in which variables of interest can be shown graphically by pull-down menus. Variables are bone and joint rotations, muscle length, muscle forces, joint reaction forces, joint moment equilibriums, muscle moment arms, muscle power and reaction forces at the articular surface of the glenohumeral joint. In addition, the output is visualized using SIMM (Musculographics, Inc.).

DISCUSSION

SPACAR allows for forward dynamic as well as inverse dynamic simulations, which have both been implemented. Forward dynamic models require too much computing power for optimization, which is not feasible in the present configuration. Inverse dynamic models predict muscle and other forces for a recorded motion. The effect of muscle dynamics can be taken into account using an inverse muscle model, which calculates the neural input to the muscles. Using this neural input for forward dynamic simulations is too simple: Instabilities will occur inevitably, i.e. due to numerical errors or due to unstable DOF, e.g. when the arm is raised above shoulder level. An efficient combined forward-and-inverse

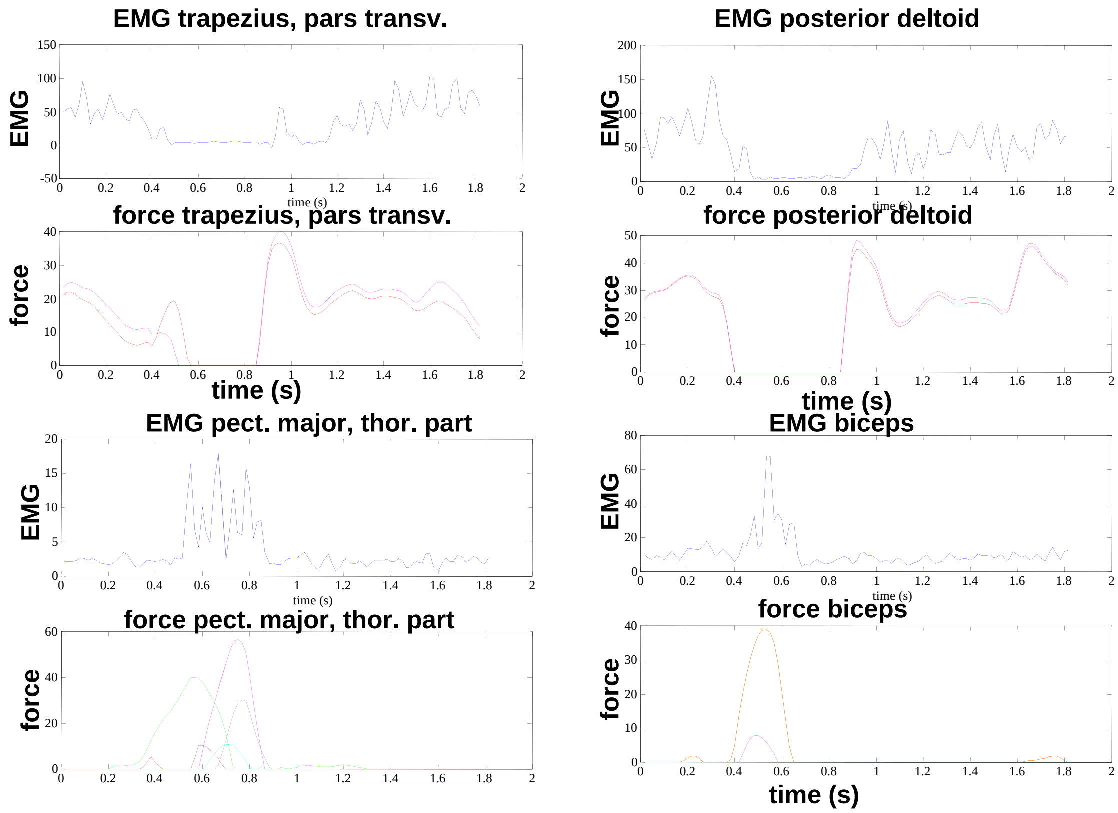

Fig. 2: *EMG to force comparison for a wheelchair propulsion cycle.*

dynamic optimization scheme, which results in a stable forward dynamic simulation, is being implemented.

The motion constraints of the STGP pose difficulties to the generation of motion equations. However, in SPACAR the system of motion equations is calculated as a function of the DOF. For the shoulder girdle only three DOF are used. This poses a related problem, since the recorded motion might not be feasible with the motion constraints. This problem is clearly evident since the cadaver morphology in the model is different from the morphology of the subject in which the motions were measured. In the DSM the input motions of the model are optimized, i.e. the recorded motions are minimally adapted such that they can be simulated in the model. Differences in the recorded and simulated motions are normally within 3-5 degrees.

An inverse dynamic simulation starts with the recorded motion of an individual, but uses the morphology of a cadaver for the model parameters. This poses a problem for a closed kinematic chain such as the shoulder girdle, since motion constraints of the cadaver prevent the exact matching of the recorded motion of the individual. A solution has been tested by morphing the cadaver geometry to the individual's geometry. The conclusion was that this procedure could not be validated, since the morphed simulations resembled the original morphology more closely than the individual's morphology.

The muscle length-tension relation is important for the correct estimate of force exertion, and is probably optimized to its function. However, the upper extremity has a wide range of functions, and it is not yet clear what functional meaning can be given to the length-tension properties.

A model is by definition a simplification of reality. Small-scale models are needed to focus on principles in motor control, large-scale models are needed for detailed predictions on the mechanical load of individual structures. Therefore, a large-scale model must have a fair detail in model structure and parameters, and must be validated. If another optimization criterion is taken, the amplitude of the muscle forces will change, but the timing patterns will hardly change. Muscle forces can not be measured directly, therefore the model can not be validated in the strict sense. Comparsions with EMG timing patterns reveal that the shoulder and elbow model has a good accuracy in predicting which muscles are active, e.g. during wheelchair propulsion (Fig. 2).