Guidance for quadriceps rehabilitation based on AnyBody

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Abstract. This paper aims to perform an objective quantitative analysis on quadriceps rehabilitation, and investigate the muscle status in cross-over design under rehabilitation actions. Rehabilitation actions are modeled using the human body modeling software AnyBody, which can analyze the variations in the muscle activity and muscle force of the quadriceps femoris during thigh flexion. In addition, it can experimentally validate the effectiveness of the model in combination with electromyographic (EMG) signals of three quadriceps femoris muscles during different activities. According to the study results, the rehabilitation actions of the quadriceps femoris can be quantified by means of collecting EMG signals.

1. Introduction

As one of the largest and strongest muscles of the human body, the quadriceps femoris is indispensable for calf stretch, thigh flexion, knee extension, hip flexion, standing, and many other daily activities, while its strength is a predictor of many lower limb functions [1, 2]. However, lack of exercise, muscle injury and arthritis [3] may cause a decline in muscle strength, thus affecting daily activities.

Resistance training and centrifugal training [4-6] are important muscle strengthening methods which have been widely used in the field of rehabilitation medicine. Regarding the quantitative features of these trainings, the real-time, non-invasive detection and interventional guidance of muscle length, feather angle, and other parameters and functions, based on ultrasound [7] and surface electromyography signals [8], have emerged as new weapons in rehabilitation.

Computer simulation software, such as ADAMS, AnyBody can simulate the biomechanical responses of human bones and muscles to environments using mathematical modeling techniques [9-11], and effectively predict the muscle status in actual experiments [12], providing some reference for rehabilitation. So far, studies on the traditional methods of evaluating the muscle force of the quadriceps femoris collect the peak torque of the muscle during rehabilitation training [13], or measure the muscle circumference after several rehabilitation trainings [14]. Rehabilitation training relies generally on the experience of the physicians; however, there is also the problem of how to choose a moderate degree of rehabilitation training in the early stage, so that it would not cause secondary injury to patients.

Furthermore, existing rehabilitation strategy introduced below cannot meet individualized parameter demands in injuries of different degrees.

In view of this, this paper selected the common recovery actions of quadriceps femoris [15] during rehabilitation as the object of study, built a rehabilitation platform using SolidWorks, and imported the model into AnyBody to simulate knee extension and hip flexion. In order to guarantee patient safety during training, it is necessary to customize the parameters of the rehabilitation platform [16]. On this basis, Matlab was used to analyze the collected EMG signals, and assess the muscle status of the
quadriceps femoris under different conditions, aiming to offer correct scientific guidance for patients with injured quadriceps femoris muscles.

2. Modeling

2.1. Muscle recruitment model

The AnyBody Modeling System (AMS) provides a complete human body system composed of bones, muscles, and nerves, which can be used for inverse dynamics analysis of the muscular system [17]. In inverse dynamics analysis, there is always the problem of redundancy within the muscular system, that is, the number of muscles comprising the muscular system is greater than that needed to balance external forces. This involves the problem of muscle recruitment. In inverse dynamics, the solution to the muscle recruitment problem is usually expressed as a problem of mathematical optimization, i.e., the minimization of the value based on the objective function $G(f^{i(M)})$, which can be expressed as:

$$G(f^{i(M)}) = \sum_{i=1}^{n^{M}} f^i_{M} / N_i^M$$

Equation (1)

$$Cf^{i(M)} = \sum \frac{d}{Cf}$$

Equation (2)

$$0 \leq f^{i(M)} \leq N_i^M i \in (1,2..n)$$

Equation (3)

In Equation (1), G is an objective function which aims to minimize all the known forces in the problem, which are expressed as muscle force $f^{i(M)}$. In addition, $f^{i(M)}$ denotes the muscle force of the $i$-th muscle and $n^{M}$ denotes the number of muscles. Equation (2) is a dynamic balance equation that is optimized as a constraint condition, where $C$ denotes the coefficient matrix of unknown forces and d represents all known loads. Equation (3) indicates that a muscle can be only pulled, but not pushed, and that its upper limit is restricted. $N_i$ denotes the muscle strength. In order to effectively enhance muscle coordination, the Hill model is adopted as the basic model for lower limbs, which is a polynomial muscle model, where the p value is set as 3 [18]. Leon et al. proved the reliability of the simplified model.

![Figure 1. Lifting platform.](image1)

![Figure 2. Human-machine coupling interface.](image2)

2.2. Coupling model

In order to faithfully simulate the movement status of the quadriceps femoris during rehabilitation and reduce errors that cannot be controlled by experimenters in the movement process, a simple lifting platform was built to assist patients in movement (Figure 1). The lifting platform is designed for the lifting of lower limbs under constant-speed and the rehabilitation actions under loads, intending to help patients train and rehabilitate their quadriceps femoris in isokinetic motion. It is mainly composed of five parts, namely, a sliding platform, an angle regulator, a pedal, a control device, and a fixing frame. The fixing frame is mounted at the top with a motor, and the middle screw is connected to the connecting plate. The connecting plate and the sliding platform are connected by T-slots and T-bolts. The angle regulator can adjust the height of the connecting plate at the screw rod according to the specific lifting...
angle of the knee joint, thereby changing the angle of the sliding platform. The pedal rises along the oblique line of the platform to complete simulation actions.

SolidWorks 2016 was used to design and transform the lifting platform model, which was imported into the AMS 6.0 simulation platform, and the AnyScript programming language was employed for modeling. Notably, before importing the model into AMS to simulate the human-machine coupling model, it is necessary to check for any redundant degrees-of-freedom, since, if there is a redundant degree-of-freedom, the model imported into AMS will face the problem of redundancy and fail to complete the designated actions. After confirming that there is no redundant degree-of-freedom problem in the model, the human-machine coupling model can be built, the constraint conditions of the human body and mechanisms can be added, and finally, rehabilitation action simulations can be performed. The coupling results are shown in Figure 2.

3. Simulation and experiment

3.1. Experimental preparations and methods

The technical framework of this study is shown in Figure 3. Clearly, the first step is to locate the positions where EMG signals should be collected from, and import corresponding human body parameters into AnyBody to perform simulations under different weight resistances. Next, a signal collector is used to collect EMG signals of the vastus medialis, rectus femoris, and vastus lateralis. The final experimental results are compared to the simulation results to validate the correctness of the simulation model.

![Figure 3. Framework of validation implementation.](image1)

![Figure 4. Muscle patch locations.](image2)

Our experimental object is a 25-year-old male, 180 cm in height, and 75 kg in weight. The subject can fully understand the meaning of this experiment, and volunteers to participate in this study. He has no severe trauma history, and no lower limb or neurological dysfunction. Considering that muscle signals are weak electrical signals [19], and that skin oils have a substantial effect on signal intensity, the skin surface at the experimental site should be wiped down with medical alcohol in advance, in order to improve the effect of EMG signal collection. It is better to evaluate the functional state of muscle by measuring a series of parameters reflecting muscle complex in the case of passive isokinetic motion than measuring muscle strength with bare hands [20]. The experimental devices included the ELONXI myoelectric apparatus, the loading device, and the lifting platform.

At the initial stage of the experiment, it is necessary to measure once the maximum voluntary contraction (MVC) of the muscle, which is defined as the weight that can be lifted only once during the isometric contraction training of muscle force. The electric discharge produced by a muscle represents its force production status. Referring to the MVC posture corresponding to each muscle recorded in the Handbook of Manual Muscle Testing [21], the electric discharge by each muscle has been collected.
using the signal collector, denoted as $S_{\text{max}}$. In the AMS system, the muscle activity is equal to the current muscle force divided by the muscle strength. Introducing it into EMG signals yields the muscle activity:

$$A_c = S / S_{\text{max}}$$

(4)

where $S$ denotes real-time EMG signals.

3.2. Collection and processing of EMG signals

The ELONXI sEMG signal detection system, developed by Jiaopu technology can record the surface myoelectric signal of lower limb movement. During the experiment, the main force producing muscle of this movement is quadriceps femoris, and the main members of quadriceps femoris are four muscles, among which the muscle closest to the skin surface is the vastus lateral muscle (VL), rectus femoris (RF), and vastus medial muscle (VM). The vastus intermedius muscle (VI) belongs to the inner layer of the quadriceps femoris and is covered by the three muscles described above, which is not easy to measure; thus, it is excluded from this study.

In order to acquire good signals, attention should be paid to the following matters [22]: (1) Electrode position: The crosstalk effect of other muscles at the electrode position should be minimized. Detection electrodes should be placed at the middle of the muscle belly, as far away from other muscles as possible. The direction of the electrode pair should be parallel to the muscle spindle direction. Reference electrodes should be placed at locations with the fewest muscles. (2) Electrode spacing: When the electrode spacing is large, the acquired signals are broader and deeper with higher amplitudes. Consequently, to ensure the comparability of measurements, the electrode spacing should be fixed. Specific muscle patch locations are shown in Figure 4.

As regards the original sEMG signals measured using surface electrodes, the first move is to provide 50 Hz notch processing to eliminate the power line interference; next, 20–500 Hz band-pass filtering is performed based on infinite impulse response (IIR) to complete the processing of the original signals [23]. After that, the sliding window method (window length=300 ms) is employed to compress the processed signals. The size of the sliding window is 50 ms. As can be known from the average EMG (RMS) equation, adopting the sliding window method exerts no effect on the calculation result of the final muscle activity. More specifically,

$$RMS = \frac{1}{n} \sum_{i=1}^{n} x_i$$

(5)

where $n$ denotes the window length and $x_i$ denotes the amplitude of the EMG signals per unit time.

4. Results and discussions

4.1. Experiment results

The action of knee extension and hip flexion under different load from 5 kg to 20 kg was shown in Figure 5. With the increase of load, the activity is increased. The angle increases for the first 2.5 seconds, and decreases for the rest of time. So the activity of rectus femoris decreased when angle increased, the activity of vastus lateralis and medialis increased when angle increased.

4.2. Comparison of simulation and experiment results

Based on the progressive resistance design results in Figure 6, with the increase of load, the activity of the muscle presents an increasing trend; with the increase of knee extension and hip flexion angle, the activity of the rectus femoris declines, while the maximum activity of the vastus medialis and the vastus lateralis increases significantly. Within one cycle, the activity of the rectus femoris first declines and then increases, while that of the vastus medialis and the vastus lateralis first increases and then declines.

The action of knee extension and hip flexion under a load of 5 kg was taken as an example, and the simulation and experimental results of muscle activity were compared (Figure 7). Clearly, there are
some errors between simulation and experimental data; however, they have presented a basically consistent trend of muscle activity. The trends under different loading conditions are the same. This demonstrates that it is reliable to introduce the concept of activity.

![Figure 5](image1.png)

(a) Rectus femoris  
(b) Vastus lateralis  
(c) Vastus medialis

**Figure 5.** Progressive load combination of muscle activity versus time in experiment.

![Figure 6](image2.png)

(a) RF muscle  
(b) VL muscle  
(c) VM muscle

**Figure 6.** Progressive load combination of muscle activity versus time.

![Figure 7](image3.png)

(a) Rectus femoris  
(b) Vastus lateralis  
(c) Vastus medialis

**Figure 7.** Comparison of simulation and experimental results on muscle activity versus time.

4.3. Ultimate load and Angle

From the definition of muscle activity knowns that the maximum value of muscle activity is 1. Proper exercise helps to strengthen the muscle while excessive exercise (activity > 1) will cause muscle injury. Figure 8 shows the variations in muscle activity with the increase at a lifting angle of 25°. It can be seen that the muscle activity of the rectus femoris reaches 100% when the resistance force is 172kg, in which case muscle injury will occur. In this process, the activity of rectus femoris accounts for the largest proportion of the total activity, thus, it can be seen as the muscle with the largest contribution.

Clearly, the forces applied on the knee joint and the hip joint within one lifting cycle are the load components along the x-, y-, and z-axis directions. According to Figure 9 and Figure 10, when the movement angle reaches about 125-126 degrees, the load component of the knee and hip joint in the y-direction have experienced abrupt variations, while those in other directions remained steady.
Figure 8. Activity-load relationship at maximum angle within one cycle.

Figure 9. Load component in the knee joint.

Figure 10. Load component in the hip joint.

Focus on the effect of angle on the load components in the y-axis direction, it can be seen in Figures 11 and 12 that the components of load in the hip joint and the knee joint in the y-axis direction begin to produce abrupt variations when the angle reaches a value of 125-126° under unloaded conditions. In order to prevent abrupt variations in force from causing joint injury, the ultimate angle in rehabilitation actions should be set as 125°.

Figure 11. Knee joint-load force curve.

Figure 12. Hip joint-load force curve.
4.4. 3D diagram of the variation rules of muscle force

Figure 13 shows the relationship of load force with hip joint angle and the time taken to complete one action. The time axis reflects the action speed and the hip joint angle reflects the action amplitude. As it can be seen in Figure 13, the variation trends of muscle force and muscle activity are basically consistent. In the action of knee extension and hip flexion, the forces applied on the rectus femoris, the vastus medialis, and the vastus lateralis are sensitive to angle variations. With the increase of angle, the force applied on the rectus femoris and its activity exhibit an obvious declining trend, while the forces applied on the vastus medialis and the vastus lateralis and their activity increase significantly. It is also clear that, based on the hip joint angle and the time taken to complete one action, the muscle activity and the corresponding muscle force under such conditions can be easily obtained.

![3D diagram of muscle force variations](image)

**Figure 13.** Variations in two-factor variable - muscle force/muscle activity.

5. Innovative points

The use of the AnyBody software by domestic scholars has been limited to the simulation of rehabilitation actions [24, 25]; however, their results have not been compared with experimentally obtained EMG signals, thus the reliability of the simulation experiment cannot be guaranteed. In this study, the introduction of EMG signals to calculate the muscle activity has effectively validated the reliability and accuracy of the simulation experiment.

Through AnyBody simulations, where the needed parameters can be input, the muscle force variations of individuals can be obtained and the suitable angle and loading conditions can be acquired, in order to exercise properly and avoid injury. That is to say, quantifying the intensity of rehabilitation actions in the early stage is conducive to reducing misjudgments by the medical staff due to lack of experience.

**Acknowledgements**

The work was supported by the National Natural Science Foundation of China (No. 51775114, No. 51275092) and Major R & D platform for basic parts technology of industrial robots in Fujian Province of China (2014H21010011).
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