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Trajectory Planning of a Robot for lower limb Rehabilitation

Y. Pei, Y. Kim, *S. Member, IEEE*, G. Obinata, *S. Member, IEEE,* K. Hase, and D. Stefanov, *S. Member, IEEE*

Abstract— **We introduce a method for lower-limb physical rehabilitation by means of a robot that applies preliminary defined forces to a patient's foot while moving it on a preliminary defined trajectory. We developed a special musculoskeletal model that takes into consideration the generated muscle forces of 27 musculotendon actuators and joint stiffness of the leg and allows the calculation of the motion trajectory of the robot and the forces that the robot needs to apply to the foot in each moment of the therapeutic exercise. Robotic treatment programs are customized for the individual patient by using a genetic algorithm (GA) that refers to the musculoskeletal model and calculates the parameters of the spline curves of the motion trajectory of the robot and forces acting on the foot.**

I. Introduction

THE increased needs of physical rehabilitation catalyzed the development of new techniques and training equipment for helping the individuals to enhance their movement abilities and to prevent the loss of mobility.

New rehabilitation techniques are enhanced further by varying selected parameters of the rehabilitation process and studying the effect of the changes on the user's improvement. Such studies often involve large patients' groups and monitoring of multiple physiological and mechanical parameters during the rehabilitation interventions [1].

Modern design of physical therapy programs often uses musculoskeletal models of the body segments and computer simulation of the exerted muscular forces as a tool for analysis of the effect of the planned physical exercises. This way, such models accelerate the process of refinement of the developed programs and facilitate the identification of the conditions that maximize the treatment effect. The development of much accurate musculoskeletal models includes research on the muscle forces [2], investigation on the response of the joints to the external load applied to them [5], exploration of the passive reaction forces and muscle forces generated during exercising in closed- and open-kinetic chains [3] , [4].

Musculoskeletal modeling and dynamic simulation have been widely used for analysis of lower limbs rehabilitation strategies. The approach has been applied successfully by many authors for designing movement trajectories that reduce the loads of the joints and maximize crank power [6] - [8].

In this paper we introduce a method for lower-limb physical rehabilitation by using a robot that applies forces to the patient's foot while moving it on the designated trajectory. We developed a special lower-limb musculoskeletal model that considers the effect of the generated muscle forces and joint stiffness and allows calculation of the motion trajectory of the robot and the forces that the robot needs to apply to the foot in each moment of the therapeutic exercise. When the robot needs to respond to artifacts due to sudden unintended patient's motions or system failure, the same model can be used for efficient modification of the robot trajectory and correction of the forces applied to the patient. We also developed a genetic algorithm (GA) for automatic generation of customized treatment programs for individual patients. The algorithm refers to the musculoskeletal model and calculates the parameters of the spline curves of the motion trajectory of the robot and forces acting on the foot.

II. LOWER LIMB MUSCULOSKELETAL MODEL

In this study we use a bipedal musculoskeletal model which consists of 7 rigid segments representing the trunk and two legs (Fig. 1). Each leg consists of 3 rigid segments representing the femur, the tibia, and foot. Each leg is connected to the pelvis which is represented also as a rigid body. The tibia and the fibula are represented as a single rigid segment.

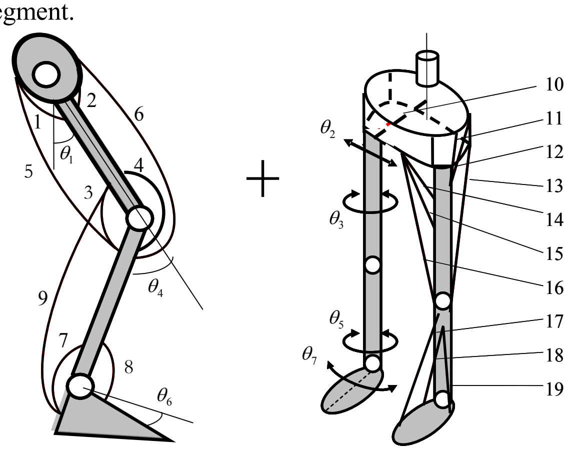

Fig. 1. Musculoskeletal model

With respect to the pelvis, each leg of the model has a total of 7 degrees of freedom which allow leg movement in 3

Y. Pei, Y. Kim, and G. Obinata are with the Graduate School of Engineering, Nagoya University Furo-cho, Chikusa-ku, Nagoya, 464-8603, phone: +81-52-789-5030; e-mail: obinata@mech.nagoya-u.ac.jp.

Kazunori Hase, is with the College of Urban Liberal Arts, Tokyo Metropolitan University, Minamioosawa 1-1, Hachioji, Tokyo 192-0397, E-mail: kazunori.hase @tmu.ac.jp..

D. Stefanov is with the Faculty of Engineering, Coventry University, Coventry, CV1 5FB, UK, e-mail: d.stefanov@coventry.ac.uk.

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dimensions. The hip and ankle joints of each leg are modeled as frictionless joints with 3 degrees of freedom. The knee joint has a moving center-of-rotation for flexion and extension, defined as a function of the knee flexion angle. For the composition of the model, in order to avoid further design complexity and at the same time to give reasonable accurate representation, we ignore or group some small muscles. Each leg of the musculoskeletal model is driven by 27 musculotendon actuators, which are combined into 19 muscle groups as shown in Fig. 1. Each group consists of one or more muscles which receive the same excitation signal. The muscle groups that power the leg motion in the sagittal plane are shown in Fig. 1a (positions 1 to 9). Fig. 1b presents the remaining 10 muscle groups that are responsible for the internal and external rotations of the hip and ankle. For simplicity, in this model we ignore the Coriolis forces which are considered to be relatively small.

In this study we refer to the muscle model proposed by Hill for evaluating muscle contraction dynamics [9]. The mass and moment of inertia for each body segment are estimated for a standard human body by the method described in [10]. Links of the model are assumed to be rigid. The relationship between the generated joint torque and the muscle forces that cause the torque can be described with the following equation.

$$\tau_i = \sum_k r_{i,k} \times f_k \tag{1}$$

where τ_i is the torque at joint i; f_k is the muscle force generated from muscle k, $r_{i,k}$ is the distance from the i-th axis of rotation to the particle of the k-th muscle force applied to the same link.

All muscles attached to the link will work together to contribute to the total generated torque of the joint. However, muscle strength is dependent on the muscle lengthening which changes depending on the angular rotation of the joint. Because of that, the balance of the acting muscle forces and the joint torque respectively cannot be determined correctly. As reported in [11], there is good correlation between the muscle physiological cross-sectional area (PCSA) and the force generated by the muscle. The force of muscle j can be estimated from the force strength index $u(f)$ after its minimization:

$$u(\mathbf{f}) = \sum_k \left(\frac{f_k}{PCSA_k}\right)^3 \tag{2}$$

where $PCSA_k$ is the physiological cross-sectional area of the k muscle and f_k is the muscle force which we want to estimate.

The minimization of $u(f)$ leads to a good match between the PCSA and actual muscle forces [11]. On the other hand, less muscle fatigue is achieved when the strength index $u(f)$ tends to be minimum.

During the joint rotation, the muscle force f_k changes within the following boundaries:

$$f_{k\min} \le f_k \le f_{k\max} \tag{3}$$

If two or more muscles with different lengths are attached to the same link at different points, the minimization of (2) can be achieved for different values. These lead a nonlinear problem for the minimization.

III. DESIGN OF THE JOINT TRAJECTORIES AND APPLIED EXTERNAL FORCES

A. Generating joint trajectories for rehabilitation training

In this study, desired joint trajectories of the leg are realized by the movement of the foot. Other words, the trajectory parameters of the foot movement should be determined as a function of the joint angles θ_i .

It it assumed that during the rehabilitation exercise the joint angles θ_i of each joint are smooth periodic functions. The joint angle can change between upper bound θ_{imax} and lower bound θ_{imin}. The time of the cycle (T) can be considered as composed from two sections. Section 1 includes the time $0 \le t \le t_{i,1}$ when the joint angle θ_i increases from θ_{imin}.to θ_{imax} . Section 2 includes the time $t_{i,1} \le t \le T$ when the joint angle θ_i decreases from to θ_{imax} to θ_{imin}..

The joint angle of each section is represented by a fifth-degree polynomial as follows:

$$\begin{aligned}\theta_{i,j} &= a_{i,j}(t-lT)^5 + b_{i,j}(t-lT)^4 + c_{i,j}(t-lT)^3 \\ &+ d_{i,j}(t-lT)^2 + e_{i,j}(t-lT) + g_{i,j}\end{aligned} \tag{4}$$

where i is the number of the joint, j stands for the time section (j =1 or 2), T is the period of the cyclic joint movement, and l is the iteration number of the cycle.

The joint angle increases during the first time interval ($0 \le t \le t_{i,1}$). It starts to decrease with the beginning of the second time interval ($t_{i,1} \le t \le T$). A smooth transition between the sections of movement is achieved if at the beginning of each new section the following initial conditions are completed: $\theta_{i,1}(0) = g_{i,1}$, $\dot{\theta}_{i,1}(0) = 0$ and the angle $\theta_{i,1}(t_{i,1}) = \alpha_i$, $\theta_{i,2}(t_{i,1}) = \alpha_i$, $\dot{\theta}_{i,1}(t_{i,1}) = \dot{\theta}_{i,2}(t_{i,1})$, $\ddot{\theta}_{i,1}(t_{i,1}) = \ddot{\theta}_{i,2}(t_{i,1})$. Under these conditions, the problem considered in this paper becomes to search α_i , T , $t_{i,1}$, $d_{i,1}$ and $d_{i,2}$ for maximizing the performance index which is defined in Section IV.

B. Design of the forces, applied to the joint

In many cases of functional therapy, better therapeutic effect can be achieved if the forces applied to the joints during the training process are big enough. On the other hand, the load on the rehabilitated joint should be kept relatively low to avoid the risk of injuries during training.

The knee joint is most vulnerable to injury during lower limb rehabilitations due to its specific structure. As the knee flexes, the femur movement over the tibia is a combination of gliding and rolling. If the shear forces applied to the knee become very high, that may cause joint damage. The excessive shear forces applied to the knee may have even worse effect to elderly people because quite often the rolling and gliding of the femur over the tibia is impeded additionally due to the looseness of the ligament or deformation of the articular cartilage. Because of their relatively low elasticity, passive elements such as ligaments can be damaged easily if the external forces applied to the knee cause high tension to them. Knee injuries can be avoided if the external forces applied to the knee during training do not cause excessive glide movement.

The force F_{ext} applied by the robot to the tip of foot causes reaction forces F_i acting on the knee joint that can be calculated from the inverse dynamics of the musculoskeletal model. The force F_i is decomposed into the compressive component F_{iR}, which tries to compress the femur to the tibia, and shear element $F_{i\tau}$ which is perpendicular to F_{iR}.

$$F_{i\tau} = F_i - F_{iR} \tag{5}$$

Injuries or pains during training can be avoided if the shear force is limited to certain value as follows:

$$\|F_{i\tau}\| \le F_{i\max} \tag{6}$$

External forces F_{ext} that the robot will apply to the foot should satisfy condition (6). The magnitude and the direction of F_{ext} are calculated by considering the rotation angles of the base coordinate around the z and y axis and minimizing $\|F_{i\tau}\| / \|F_i\|$. The initial external force is set to $F_{ext,0}$ and we use the following formula to update F_{ext} until condition (6) is satisfied:

$$F_{ext,w} = F_{ext,w-1} + (\|F_{i\tau,w-1}\| - F_{i\tau\max}) \frac{F_{i\tau,w-1}}{\|F_{i\tau,w-1}\|} \cdot \frac{F_{ext,0}}{\|F_{ext,0}\|} \tag{7}$$

where w is the number of updating.

IV. PERFORMANCE INDEX AND SEARCH ALGORITHM ON THE PARAMETERS

A. Performance index

Even if the trajectory of the tip of foot and the force acting on it are determined, there are many joint trajectories and muscle tensions which correspond to the given trajectory and the force since there exists redundancy in the degree-of-freedom of musculoskeletal system. The level of motion jerk is an important performance index and its minimization leads to smooth joint trajectory. The joint torque value is another important performance index, and its limitation leads to smooth muscle forces.

The index of jerk can be expressed on the following way:

$$E_\theta = \frac{1}{2}\int_{t_0}^{t_f} \sum_{i=1}^{7} \left(\frac{d^3\theta_i}{dt^3}\right)^2 dt \tag{8}$$

The index for joint torque is

$$E_\tau = \frac{1}{2}\int_{t_0}^{t_f} \sum_{i=1}^{7} \left(\frac{d\tau_i}{dt}\right)^2 dt \ . \tag{9}$$

The amount of loading of the target muscles should be dosed correctly. The effect of the muscle loading can be evaluated with the integral of the generated muscle forces generated over the training time:

$$E_f = \int_{t_0}^{t_f} \sum_k f_k dt \tag{10}$$

For the design of the movement trajectory of the robot and the external force applied at the tip of foot we use an index which is the sum of the three indices. The index E is given by the following equation:

$$E = E_f - k_1 E_\theta - k_2 E_\tau \tag{11}$$

After selecting the weight coefficients k_1 and k_2, index (11) should be maximized with respect to α, t_1, T and F_{ext}. We apply a genetic algorithm (GA) to search those parameters because the optimization problem is nonlinear.

V. MODEL CALCULATION EXAMPLE

A. Constraints

Quadriceps femoris includes four muscles heads (Rectus femoris, Vastus lateralis, Vastus medialis, and Vastus intermedius). It is the great extensor of the knee joint and stabilizes the patella and the knee joint. Due to the action of the Rectus femoris, it is also a flexor of the hip. The physiotherapy restoration of the Quadriceps after injury should be done with the application of very limited load to the knee joint if the latter is also damaged. In addition, it is desirable if the ventral hip muscles generate limited tensions during the therapy exercise. These requirements can be met by limiting the range of the hip joint. We design such attainable movements by limiting the upper bounds for the angular velocities of the joints. We set the bounds as shown in Table 1.

Table I Constraints in parameters search

Number of joints to rehabilitate	$m = 2$
Maximum shear force at the knee joint	$F_{4\tau\max} = 25[N]$
Rotation range of hip joint	$-90 \le \theta_1 \le 0$[deg]
Rotation range of knee joint	$0 \le \theta_4 \le 130$[deg]
Angular velocity range of each joint	$-180 \le \dot{\theta}_i \le 180$[deg/ s]
Training cycle	$0 \le T \le 10[s]$

The rehabilitation exercise of the quadriceps femoris should exclude large passive loads on knee joints.

During the physical exercise patient is asked initially to stay off their feet and to be face-up. The forces and joint angles that correspond to the patient posture are entered to the program and the GA starts to search parameters of the joint motion that reduce the passive forces at the knee joint.

In this example, we refer to the parameters of a standard adult person with height 175cm and weight 70kg, the values of PCSA of each muscle were taken from the data published in papers [12], [13]. The length, mass, and inertia moment of each body segment are estimated by using the regression formula of Ae et. al [10].

The parameters of the GA are set as follows: the number of individuals is 100, the crossover fraction is set to 0.25, the amount of mutation is set to 0.01, the threshold value for selecting a new generation is 5% lower than the best fitness value of the generation. The search process is set to terminate if the variation of the evaluation index E (fitness function) in the gneration is lower than the threshold value for 5 consecutive generations. In this example, the values of the parameters are expressed with 7 bit binary strings.

B. Calculation results

The calculation results are shown in Fig.2. The trajectories of the joint angle, angular velocity and angular acceleration are shown in (a), (b) and (c), respectively. Parameters vary smoothly within the bounds. The applied external force is shown in (d). The shear force $F_{4\tau}$ acting at the knee joint, is shown in (e). It varies within the predefined bounds. The generated muscle forces of rectus femoris and vastus are shown in (f). These calculation results show that all constraints on the motion are satisfied.

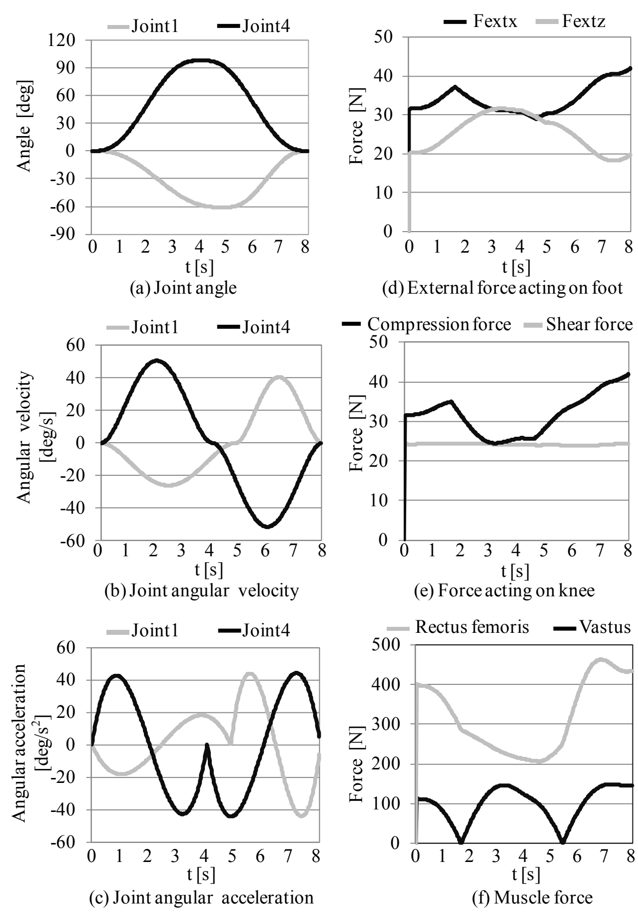

Fig. 2 Calculation results for angle trajectory, angular velocity, angular acceleration, external force, load at knee joint, and muscle forces.

Fig. 3 shows the foot motion caused by the optimised external force. The lower limb is presented as a stick figure. The vectors of the velocity and forces applied to the foot for the sequential moments of the flexion and extension are shown in the same figure. During the flexion phase, the external force assists the lifting of the limb and causes small shear force to the limb. At the beginning of the extension phase the external force continues to cause shear force to the limb because the direction of the force at that time is opposite to the limb motion (see the upper part of Fig. 3(b)). In later stage of the extension phase, the designed external force works to cancel the gravitational force acting on the limb (see the lower part in Fig. 3(b)).

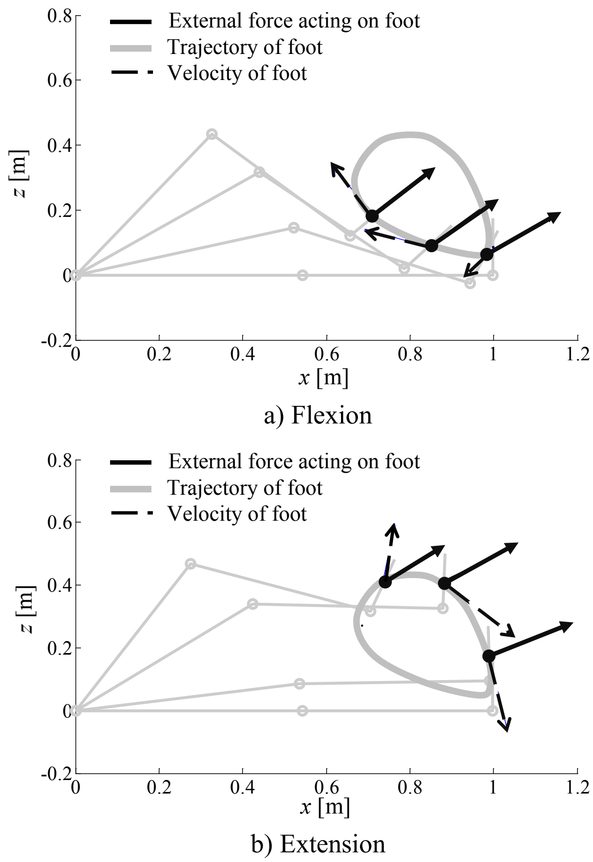

Fig. 3. Stick figure, external force, and velocity of the tip of foot for the optimized motion.

C. Comparison of the designed movement with the movements which are usually used by PT

In order to explore the effectiveness of our proposed method, we compare the results with the conventional method for training quadriceps femoris, which is usually used by PT in hospitals. The conventional procedure includes the following steps:

Step 1: The attitude of patient is initially set to stay off one lower limb for the target of the training and to be face-up, and the other lower limb is set to the position of drawing his knee up. The patient keeps the other lower limb at the initial position during training.

Step 2: The patient lifts the lower limb for the target of training up to the position such that the both femurs are at the same level from the floor while keeping the knee joint extended.

Step 3: The patient puts slowly down the lower limb for the target of training to the initial position.

The patient is asked to repeat the motions of Step 1 to Step 3. We assume that the one cycle of motion takes 4 seconds.

To obtain the moments and muscle forces generated at the joints we apply the inverse dynamics approach to the training motion and refer to the developed musculo-skeletal model. We compare this conventional method with the proposed method by using the following efficiency index of the training:

The total energy consumption of the lower limb during the time interval $t_0 \le t \le t_f$, which equals the generated kinetic energy of the corresponding rigid links in the musculo-skeletal model, can be calculated as follows:

$$E_\tau = \int_{t_0}^{t_f} \sum_{i=1}^{7} \left| \tau_i \dot{\theta}_i \right| dt \tag{12}$$

The generated energy of the target muscles in the lower limb is given by the following equation.

$$E_m = \int_{t_0}^{t_f} \left| f_k r_{i,k} \dot{\theta}_i \right| dt \tag{13}$$

We can define an index of the training by taking the ratio of these energies.

$$\eta = E_m / E_\tau \tag{14}$$

This index is the ratio of the consumed energy of the target muscles to the total kinetic energy of the movements of lower limb. In other words, the index expresses the efficiency of the designed training. The higher index value means less load energy on the regions of disorder per total consumption of energy by the limb movement.

The maximum shear force at the knee joint of the designed movements is 25% less than the maximum shear force of the conventional movement. The efficiency of the conventional movement is 21% of index (14). On the other hand, the efficiency of the designed movement is 28%. These results indicate the usefulness of the proposed method.

VI. CONCLUSION

In this paper, we introduce a method for lower-limb rehabilitation by using a rehabilitation robot connected to the foot. During the physiotherapy session, the trajectory of the limb joints and the parameters of the external forces acting to them are set by the movement of the robot and the forces that it applies to the foot. We developed a special musculoskeletal model that takes into account the generated muscle forces of 27 musculotendon actuators and the leg joint stiffness. The model is used for the calculation of the motions and forces of the robot that will cause joint motions and forces as needed for the rehabilitation process. The joint movement trajectory and external forces can be adapted easily to the patient characteristics because the method is model-based and allows changing of the parameters of the musculo-skeletal model. The proposed system allows the execution of complex 3D motions which parameters can be adapted to the particular patient to provide safety 3D exercise.

In this study we assumed that the joint angles change smoothly within the range of motion. This led to significant simplification of the optimization task. The calculation of the parameters of the training trajectory and external force exerted by the robot is a non linear optimization problem. Genetic algorithm was successfully used for that.

Finally, we proposed a post-evaluation index that allows comparison of the method for rehabilitation training. The index is the ratio between the energy needed for exercising of the target muscles and the total kinetic energy used for the movements of lower limb. We used this index to compare the efficiency of different approaches for physiotherapy. It was shown the proposed method is much efficient in the lower limb training in comparison with other conventional method used in the PT practice.

The development of a robotic system is in progress. The system will include 6 component force sensors to adapt better the motion velocity, position, trajectory, etc to the emotional and current physical state of the user.

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