Influence of limb configurations and stimulation parameters on knee and ankle joint torque generation

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Abstract

An important question in lower extremity functional electrical stimulation (FES) applications is how much does the muscular-skeletal system modulate the joint torque generation and how much has to be performed by the neuronal system to generate a normal walking pattern. In the presented study we quantified the influence of five limb configurations occurring at 7%, 45%, 65%, 75%, and 98% of the stride on the ankle and knee joint torque generation. Transcutaneous electrical stimulation was used to generate the same level of neuronal activation in the ankle and knee joint actuators for each configuration and isometric ankle and knee joint torques were measured. The results from nine healthy subjects showed that the limb configuration had a strong influence on the generated joint torques. Inter subject variations were surprisingly low. The hamstrings and gastrocnemius muscles produced high joint torques for a given electrical stimulation intensity at stride positions where the EMG activity is high during the gait cycle. This suggests that the neuronal effort is supported by the muscular-skeletal system for these muscles. For limb configurations that represent early stance phase the quadriceps and tibialis anterior muscles did not produce high joint torques as could be estimated from EMG activity.

1. INTRODUCTION

Stimulation patterns for FES walking that generate a smooth and correctly timed activation of the most important lower extremity muscles (e.g. the knee and ankle flexors and extensors) can be derived from EMG data as published by [1]. Such approaches were suggested by different authors [2, 3]. Practical applications of neuroprostheses for walking, for example the Parastep system [4], showed that it is possible to use very simple rectangular or trapezoid shaped stimulation patterns to generate movement, although the movements appeared jerky. One reason why such simplified stimulation patterns are sufficient to generate walking movements could be due to a joint angle dependent modulation of the generated torques performed by the muscular-skeletal system which reduces the neuronal effort. In the presented study we investigated the influence of the muscular-skeletal system on the knee and ankle joint torque generation. We replaced the neuronal effort with repetitive electrical stimulation at constant intensity and used the Multi Moment Chair (MMC) developed by the Group of Riener in Munich [5] to measure under isometric conditions the influence of limb configurations to the muscle torque output.

2. METHODS

2.1. Subjects and Setup

Nine healthy subjects (4 male, 5 female, age 37.3 ± 15.6 years) were selected for the study. The subjects reported to be average sporting (5.7 ± 2.1 on a scale of 1-10) and had no lower limb injuries in the past 6 years.

The subjects were seated in the MMC with the upper body fixed by straps over the shoulders and the waist. The left foot was rigidly attached to the MMC's foot holder, which was instrumented with a 6 DOF load cell from ATI. Under isometric condition the hip, knee, and ankle joint torque could be calculated from the measured forces and torques at the rigidly attached foot. In the presented study we were interested in the knee extension (KE), knee flexion (KF), ankle plantarflexion (APF), and ankle dorsiflexion (ADF) joint torques that were generated by electrical twitch activation of the quadriceps (QC), hamstrings (HS), gastrocnemius (GM), and tibialis anterior (TA) muscle groups.

2.2. Stimulation parameters

5x5 cm (for TA and GM) and 10x5 cm (for QC and HS) self-adhesive transcutaneous stimulation electrodes were placed over the motor points of the muscles. A Compex Motion electric stimulator [6] provided current regulated monophasic pulse trains for two stimulation patterns with equal charge of either eight 250 µs pulses @ 100 Hz or two 1000 µs @ 25 Hz with a train duration of 80 ms each. With the first pattern we aimed to stimulate more the efferent fibers (short pulse duration) and with the second afferent fibers were also activated. The stimulation amplitude was adjusted to produce joint torques of 10% of maximum voluntary contraction (MVC) for all muscle groups except for the TA where we experienced over-stimulation and fatigue and therefore reduced the intensity to 5% MVC. The stimulation trains were consecutively applied to the four muscle groups every 3 s, such that each muscle group received an 80 ms stimulation twitch every 12 s. This guaranteed non-fatiguing stimulation during the entire experiment.

2.3. Protocol

The isometric MVC joint torques KE, KF, APF, and ADF were determined in a sitting position with hip 65º, knee 62º, and ankle 0º (see lower right line drawings in Figure 1). In this position the stimulation amplitudes were then set for both trains to generate knee and ankle joint torques of 10 %, respectively 5 % of the MVC torques.

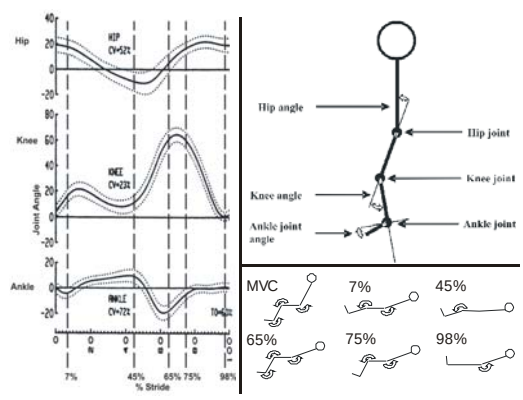

Figure 1: Hip, knee, and ankle joint angles were derived from kinematic data for normal walking [1]. The line drawings on the bottom right show the actual positions in the MMC.

During the experiment the subjects' left lower limbs were randomly adjusted to five limb configurations typically observed during normal walking at 7 %, 45 %, 65 %, 75%, and 98% of the stride. In each configuration the two stimulation trains were both randomly repeated for 10 times for each muscle group. Table 1 lists the adjusted hip, knee and ankle joint angles, and the stimulation parameters. Figure 1 shows the chosen normal walking trajectories from Winter [1] (left), and with line drawings (right bottom), how the subjects were positioned in the MMC, rotated at ~90º with respect to the real walking situation.

% stride	7%	45%	65%	75%	98%
Joint angles	hip 17° knee 17° ankle -4°	hip 0° knee 17° ankle 9°	hip 17° knee 62° ankle -18°	hip 17° knee 62° ankle -4°	hip 17° knee 0° ankle 0°
Quadriceps	Amplitude: generating 10% MVC-torque (TA: 5% MVC)				
	Inter-train interval: 12 s				
Hamstrings	Current regulated waveform: monophasic				
Tibialis Anterior	Trains: 10 * at 250 µs pulse width, 100 Hz, 8 pulses 10 * at 1000 µs pulse width, 25 Hz, 2 pulses				
Gastro-cnemius	Randomization of the two stimulation trains and Randomization of the % stride (trials)				

Table 1: The joint angles for the five limb configurations and the stimulation parameters and patterns for the four muscles are listed.

3. RESULTS

Isometric torque twitch responses generated by 80 ms stimulation pulse trains in the knee and ankle joints were measured using the MMC. The torque responses were recorded with 100 Hz sampling frequency. The following torque peaks were used for the analysis: KE torque for QC stimulation, KF torque for HS stimulation, APF torque for GM stimulation, and ADF torque for TA stimulation. For each muscle and subject the peak torques were normalized to the maximal generated peak torque of the muscle over all limb configurations. The mean value of the ten twitch responses of the same condition was taken for further analysis. Figure 2 shows the mean and the standard error of all nine subjects for each muscle, the two stimulation trains and the five limb configurations. The small standard error indicates very good reproducibility of the twitch responses and the absence of muscle fatigue, since the different limb configurations were randomized among the subjects as well as the two different pulse trains. It could be observed that in contrast to the sitting position the 1000 µs pulse trains produced less torque in most other limb configurations than the 250 µs pulses. The differences in the generated torques

were between 5 % and 16 % and were statistical significant ($p<0.05$), except for the HS for all limb configurations and the GM for 98 % of the stride (heel strike). An exception could be observed for the TA muscle: At the beginning of the swing phase (65 % of the stride) where TA starts to become active for foot lifting the 1000 µs pulse train produced a significantly (13.6 %, $p<0.01$) stronger torque than the 250 µs pulse train.

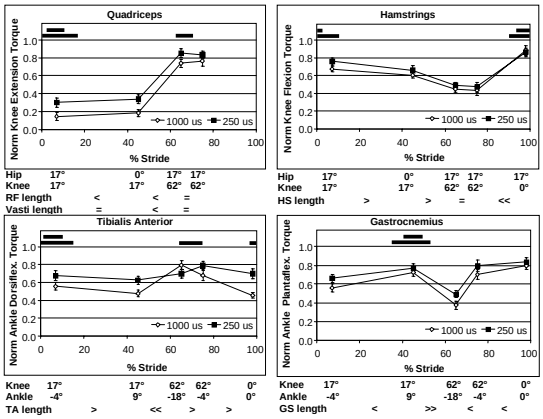

Figure 2: The four panels show the normalized means and standard errors of the stride position dependent ankle and knee joint torques generated by electrically stimulated 80 ms twitches in 9 subjects. EMG activity from [1] is depicted as black horizontal intensity bars.

The torques generated for the same neuronal input (same stimulation intensity) for different limb configurations changed considerably. For example, when the limbs were configured as in stance phase, the QC muscles produced only one third of the torque they produced in swing phase configurations. The HS and GM muscles, and the TA muscle for the 1000 µs stimulation trains, changed up to 100 % between the different limb configurations. In Figure 2 the four panels show the differences in produced muscle torque between the five limb configurations. In the heel strike position (98 %) the QC were not stimulated because the knee joint was in its fully extended position and unable to produce an extrinsic torque. On the top of each of the four panels in Figure 2 the EMG activity of normal walking derived from [1] is indicated with two vertical bars over the stride (1 bar = medium EMG activity, 2bars = strong EMG activity). On the bottom of each panel the joint angles for the five limb configurations and the change of passive muscle length between the configurations are indicated.

4. DISCUSSION AND CONCLUSIONS

We could observe a strong correlation between the passive muscle length and the generated muscle torque for all limb configurations and muscle groups that can be explained by [7]. For HS and GM the EMG activity (and indicator of produced muscle force) is high when the torque generation in these muscles is most efficient. For these muscles the neuronal effort (control effort) can remain low since the modulation of the muscle activity is also produced by the muscular-skeletal system. QC and TA produce high EMG activity at the beginning of stance phase when these muscles do not produce the strongest torque for a given neuronal activity. Nevertheless in these muscles it could also be observed that the muscular-skeletal construction supports the EMG modulation (see 65 % of the stride for QC and 65 % and 75 % of the stride for TA).

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