INTRODUCTION

Functional electrical stimulation (FES) has been explored as a means of restoring lost function in the spinal cord injured (SCI). Specific training with FES can cause significant improvements of the cardiovascular and pulmonary systems (Figoni et al., 1990; Janssen et al., 1998; Pollack et al., 1989), reduce atrophy of skeletal muscle, increase lower limb circulation and improve immune system function (Janssen et al., 1998; Ragnarsson et al., 1988), reduce edema (Faghri, 1997), increase bone density (Moor et al., 1997) and also lead to psychological benefits (Janssen et al., 1998; Ragnarsson et al., 1988). The most common type of FES is surface stimulation with the electrodes attached to the skin (Brennen, 1976).

Muscles have a high ability to adapt to altered functional requirements, and the severely reduced muscle activity after spinal cord lesion leads to muscle atrophy and changes in muscle structure. Fiber distribution in paralysed muscle is transformed towards a predominance of fast contracting type II muscle fibers (Andersen et al., 1996). Consequently, paralysed muscles behave differently than muscles in neurologically intact limbs. Additionally the force output of an electrically activated muscle is influenced by stimulation frequency and intensity (Watanabe et al., 1999; Gorman and Mortime, 1983; Durfee and MacLean, 1989), muscle fatigue (Solomonow, 1984) and other parameters like the placement of the electrodes on the skin and tissue electrical properties (Grill, 1999).

Biomechanical models have been used to investigate single- and multijoint movements (Ferrarin et al., 2001; Lan et al., 1991) as well as more complex movement patterns as standing (Khang and Zajac, 1989), walking (Popovic et al., 1989) and cycling (Gföhler and Lugner, 2000) by means of FES. In order to model muscle behavior it is necessary to determine the mechanical properties of the electrically stimulated paralysed muscles. Gerrits (2000) studied the contractile properties of the paralysed quadriceps muscles. Sinclair (2001) has done detailed investigations on the characteristics of paralysed quadriceps muscles activated by monophasic constant voltage stimulation and included the results in a simulation of FES cycling. Rochester et al. (1995) determined the contractile properties of electrically stimulated tibialis anterior muscle. Stein et al. (1992) investigated the effect of electrical stimulation training on the characteristics of tibialis anterior muscle.

The purpose of this study was to determine the mechanical characteristics of leg muscles which are artificially activated by biphasic constant current stimulation and to identify all parameters which are necessary for describing the mechanical muscle behavior in a Hill-type muscle model. The characteristics of quadriceps and hamstring muscles, which are the main actuators in FES applications for the lower limbs, were determined by a series of isometric and isokinetic measurements on an isokinetic dynamometer over the knee flexion angle range from 10 to 120°. Behavior of electrically stimulated muscles in able-bodied and paraplegic subjects was compared in order to identify the characteristics that are due to the artificial activation only and the ones due to the structural changes in paralysed muscle.

METHODS

Five neurologically intact and five paraplegic subjects were recruited for the measurements on an isokinetic muscle dynamometer (Cybex norm 2). The study was approved by the ethics commission and all test subjects gave written informed consent.

Figure 1: Leg of paralysed test person mounted on the dynamometer and schematic of the measurement system (here only the electrodes for stimulation of the quadriceps muscles are shown).

A series of isometric and isokinetic tests was done with quadriceps and hamstring muscles, which were stimulated via surface electrodes with biphasic rectangular constant current pulses. Pulse width was held constant at 400 µs.
The stimulation frequency was 40 Hz except for the tests where knee torque as a function of stimulation frequency was investigated. The applied knee torque was measured by the dynamometer and the raw signals of knee torque, knee flexion angle, and knee angular velocity were collected and stored in the notebook at a sampling rate of 500 Hz for calibration and offline analysis.

Figure 1 shows a paraplegic test person’s leg mounted on the dynamometer and a schematic of the established measurement system. The test person was seated on the dynamometer’s chair in an upright position with 90° hip flexion angle and the axis of the knee joint adjusted to the dynamometer’s axis of rotation. To avoid non-reproducible time delays a constant stimulation signal was sent out of the stimulator and the time interval where each muscle was stimulated was controlled by the computer via relays which interrupted the signal between stimulator and electrodes.

At the beginning of each measurement surface electrodes were attached to the skin over quadriceps and hamstring muscles and the stimulation current for the tests was determined for each muscle group. The current was set so that maximal muscle force was generated without apparent activity of antagonistic muscles.

1.2 s CH 1quad 1.2 s CH 2 ham 1.2 s
7 s 7 s 7 s

Figure 2: Stimulation sequence for isometric measurements.

For all isometric measurements the muscles were stimulated in the sequence shown in Figure 2. Isometric torque was measured within the joint angle range from 10 to 115 degrees knee flexion angle at 15° steps. Subsequently, isometric measurements were done at 85° knee flexion angle while current amplitude and frequency of the stimulation signal were varied. Stimulation bursts lasted 1.2 s and were interrupted by resting periods of 15.2s for each muscle. From each recorded torque curve over one stimulation interval the maximum active torque was derived by subtracting the passive torque (torque before the start of the stimulation averaged over 0.1s) from the total torque during the stimulation burst (again averaged over 0.1 s). 0-70% rise time was defined as the time interval from start of torque increase to 70% of maximum active torque. 100-30% fall time was defined as the time interval from end of stimulation to 30% of maximum active torque. The maximum active torques of the two stimulation intervals of each muscle were averaged.

Figure 3 shows the synchronization of dynamometer movement and muscle stimulation for the isokinetic measurements at joint angular velocities of 30, 60, 90, and 120°/s. At each angular velocity at first one passive up- and down movement of the leg was recorded, during the second up- and down movement quadriceps and hamstring muscles were activated in their concentric ranges. Pre-stimulation of the muscle started at least 0.6 s before the dynamometer began to move. The measured torque during the first passive up- and down movement was subtracted from the total torque during the movement with stimulation to get the active torques that were generated by active muscle forces. Resting periods of several minutes were allowed before each increase of velocity in order to reduce fatigue.

RESULTS

Figure 4 shows the measured 0-70% rise and 100-30% fall times for quadriceps and hamstring muscles of paraplegic and able-bodied subjects.
Figure 4: Comparison of 0-70% rise and 100-30% fall times and average deviations for isometric contractions of quadriceps (a,b) and hamstring muscles (c,d) as a function of the knee flexion angle. The results are averaged over all subjects of each test group. At 115° two measurements were done to get information on fatigue.

Rise and fall times were derived from the corresponding isometric torque curves and averaged over the test subjects of each group. The curves for rise and fall times over knee flexion angle of paralysed quadriceps muscles are comparable to those reported by Sinclair (2001).

Table 1: Averaged rise and fall times for isometric contractions of quadriceps and hamstring muscles for both test groups averaged over the full knee angle range.

<table>
<thead>
<tr>
<th></th>
<th>0-70% rise time [ms]</th>
<th>standard deviation [ms]</th>
<th>100-30% fall time [ms]</th>
<th>standard deviation [ms]</th>
</tr>
</thead>
<tbody>
<tr>
<td>quadriceps paraplegic</td>
<td>118</td>
<td>40</td>
<td>77</td>
<td>6</td>
</tr>
<tr>
<td>neurologically intact</td>
<td>141</td>
<td>39</td>
<td>76</td>
<td>12</td>
</tr>
<tr>
<td>hamstrings paraplegic</td>
<td>172</td>
<td>37</td>
<td>105</td>
<td>40</td>
</tr>
<tr>
<td>neurologically intact</td>
<td>264</td>
<td>79</td>
<td>107</td>
<td>28</td>
</tr>
</tbody>
</table>

Table 1 shows the averaged rise and fall times and standard deviations for isometric contractions of quadriceps and hamstring muscles for both test groups averaged over the full knee angle range. As expected due to the transformation in muscle fiber composition in paralysed muscle the rise times were longer for the neurologically intact muscles at all tested angles and for both muscles except for the hamstring muscles at 10°. In paralysed muscles fall times were shorter in the middle of the knee angle range for both muscles, in the outer regions paralysed muscles had longer deactivation times. The results show that muscle contraction induced by surface stimulation has a significantly shorter deactivation time than physiologically activated muscle.

Both rise and fall times showed no significant correlation with stimulation frequency or intensity, though for some subjects a slight increase of rise time with increasing frequency was observed.

The normalized torque-angle curves resulting from the isometric measurements are shown in Figure 5. Each subject's torque-angle curve was normalized by its peak value, then the curves of each test group were averaged. It can be seen that for both quadriceps and hamstring muscles the knee flexion angle where average maximum torque is generated is different for paraplegics and able-bodied subjects. Assuming that muscle moment arms are the same for both groups this indicates that the optimal muscle lengths are different. The quadriceps muscles of neurologically intact subjects generate their average maximum torque between 40 and 55° knee flexion angle, the torque decreases thereafter and increases again between 100 and 115°. Paralysed quadriceps muscles generate their average maximum isometric torque at the biggest knee flexion angles, but only two of the five paraplegic test subjects had the maximum torque at 115°, the others between 40 and 70°. The hamstring muscles of neurologically intact subjects generate their maximum torque between 55 and 70° knee flexion angle, whereas paralysed hamstring muscles at about 25°.

For the quadriceps muscles the average of the maximum torques of all subjects of each test group was 32 Nm for the paraplegic and 14 Nm for the able-bodied subjects. The maximum torque applied by a paraplegic subject was 87 Nm. For hamstring muscles the average maximum torque was 21 Nm for paraplegic and 7 Nm for able-bodied subjects.

Figure 5: Comparison of averaged isometric knee torque as a function of knee flexion angle and average deviations. a) quadriceps, b) hamstring muscles.
Figure 6: Comparison of averaged knee torque as a function of stimulation frequency and average deviation for both groups of test subjects at knee flexion angle 85°. a) quadriceps, b) hamstring muscles. At 50 Hz only data from three able-bodied and two paraplegic subjects was available.

Figure 6 shows the influence of the stimulation frequency on the isometric knee torque. Each subject’s curve was normalized by its value at 40 Hz, then the curves of all subjects of each group were averaged. The generated torque is increasing with increasing stimulation frequency. The increase is steeper for paralysed muscles except for hamstring muscles between 40 and 50 Hz. This may also be explained by the predominance of fast type II muscle fibers in paralysed muscles. Slow type I muscle fibers reach their maximum force at about 30 Hz, whereas fast type II muscle fibers increase their force up to 100 Hz. The results are in good agreement with data reported in the literature (Watanabe, 1999).

The influence of the knee angular velocity on the joint torque is shown in Figure 7. The dynamometer’s arm was moved in the range between 10 and 120° knee flexion angle for each subject, but as acceleration and deceleration of the dynamometer to the angular velocity 120°/s needed 40° and 15° respectively, only the resulting torques in the angle range between 25 and 85° for quadriceps and 40 to 100° for the hamstring muscles were considered at all angular velocities. Out of the measured torque-angle curves at each velocity the torques at the knee angles used in the isometric measurements (10-115° at 15° steps) were derived and normalized by the subject’s peak value at this knee flexion angle. The resulting normalized torque-velocity curves were averaged over all knee flexion angles for each subject to get one subject-specific torque-velocity curve. These curves were then averaged over all subjects of each group.

Figure 7 shows that the averaged torque-velocity curves for the quadriceps muscles are quite similar for paraplegic and able-bodied subjects. Both show an increased torque at the knee angular velocity 90°/s. For hamstring muscles the curve representing the able-bodied subjects shows higher fluctuations, again there is an increase of torque at 90°/s. The negative torque values and high average deviations might indicate that some of the able-bodied subjects were not sufficiently relaxed and therefore have to be interpreted with caution.

DISCUSSION

The muscle characteristics of electrically stimulated quadriceps and hamstring muscles in paraplegics and able-bodied subjects were determined.

One difficulty in interpreting the results is the voluntary component in case of stimulation of neurologically intact subjects. It was tried to reduce this component to a minimum by customizing the test persons to the artificial stimulation. However, a comparison of rise and fall times shows little difference for artificially and physiologically activated muscles. This indicates that the voluntary component was very small at least in the isometric measurements. The isokinetic measurements for hamstring muscles might have been influenced by voluntary muscle activity more strongly.
The reason for greater torques in the paraplegic subjects was that current amplitude applied was only 21-42 mA for the able-bodied and 55-120 mA for the paraplegic test persons. For able-bodied subjects the torques generated by electrical stimulation were between 5 and 10% of their maximum voluntary torques.

Measurements of both isometric rise and fall times and torque as a function of stimulation frequency indicate that the paralysed muscles have a higher percentage of fast contracting type II muscle fibers.

The optimal joint torques where the muscles generate their maximum isometric torques are shifted for paraplegic muscles in comparison to muscles in neurologically intact limbs. It is surprising that the optimal angles are shifted to regions where the muscles are more stretched.

The results are certainly influenced by the different training conditions especially of the paralysed muscles. Only one of the paraplegic subjects who participated in this study was very well trained with FES on a regular basis.

We hope that the results of this study will be implemented into muscle models built to simulate movements in spinal cord injured people.

**SUMMARY**

Isometric and isokinetic measurements on a muscle dynamometer were performed to determine the characteristics of leg muscles activated by surface stimulation. Measurements on the leg muscles of able-bodied and paraplegic subjects were performed to show which characteristics are due to the artificial activation only and which are due to the structural changes in paralysed muscle. Based on the results all parameters for a mechanical muscle model for artificially activated muscle can be determined.

**REFERENCES**


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