Non-repetitive Stimulation of the Common Peroneal Nerve

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Abstract

A typical pattern of stimulation for the common peroneal nerve consists of a ramp up period followed by a constant middle section and a ramp down in activity. The central part maintains a constant ankle torque to lift the toe by stimulating at a constant frequency. In contrast the natural activity of the tibialis anterior muscle varies during the gait cycle. A closer match of the torque during stimulation to the natural waveform will result in more time spent with the body weight supported by the heel as opposed to the toes. In anticipation of using non-repetitive pulses to modulate the torque during stimulation a programmable controller has been developed to investigate the effects of different stimulation profiles on normal and stroke subjects. Doublets and optimum series show enhanced torque which may lead to waveform shaping. Average torque is essentially independent of the inter-pulse duration of doublets.

1 Introduction

When a motor unit is stimulated with a low-frequency train of pulses, the addition of an extra pulse in quick succession (doublet) can significantly increase muscle tension [1][2]. Under optimal conditions, the additional tension from a doublet lasts for several seconds and greatly exceeds the tension produced by a single pulse. Studies using the 'catch effect' have suggested a slower rate of muscle fatigue [2]. Rapid muscle fatigue is the fundamental imperfection of FES, hence this result would have important ramifications in the clinical setting. Research into the catch effect indicates many results and hypothesis's supporting these findings [1]. As yet there appears no established explanation of this phenomenon. We therefore propose to exploit this effect to modulate the output from the anterior tibialis by adding "stimulation doublets" at heel rise and heel strike. Figure 1 shows the typical activity for this muscle.

2 Methods

An Odstock Dropped Foot Stimulator (ODFSIII) was modified using a PIC16F84 microcontroller to allow for the programming of different stimulation outputs and to study the response of subjects.

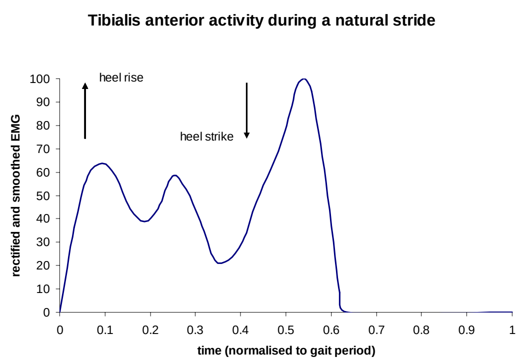


Figure 1

Ethical approval for the experimental work was obtained from the Salisbury and South Wiltshire Local Research Ethics Committee. By adjusting the stimulation intensity using the standard ODFSIII profile, an appropriate level was found for producing a comfortable foot-lift during walking. The subject was then positioned in an apparatus that measures isometrically lower limb joint moments at the ankle, knee and hip; the multi-moment chair [3]. The subject sat within the apparatus, with the feet secured in boxes mounted on beam structures with strain gauge transducers. The knees are secured in vertical support beams, again with strain gauges attached. These sensors measure any force change produced about the hip, knee and ankle in the three orthogonal planes. Sensor inputs are sampled at 100.8Hz and averaged over 16 samples. Straps around the pelvis, waist and body are used to minimise movement and for safety. Different joint angles are accommodated by inclining the back-board, raising or lowering the seat height and adjusting the angles of the feet boxes in the sagittal plane. In these tests, the subject was semi-reclined with the position defined at the ankles 65 degrees, at the knees 15 degrees and at the hips 35 degrees. This position was selected to mimic the 'extended' posture adopted during walking and for comfort.

In this study, the responses only at the ankle are investigated. By the nature of the design of the multi-moment chair, dorsiflexion moments are measured directly as one of the bending moments of the beam structure, with an associated non-systematic RMS error of 0.86 Nm [3]. Steps were taken to minimise further errors by fixing the foot securely to prevent any movement and to allow for rests in between stimulation tests to minimise the effects of fatigue. During initial measurements, it was observed that the dorsiflexion moments were only a few Nm maximum. It was therefore decided to increase the stimulation levels to result in increased moments up to about 10 Nm.

Dorsiflexion force was determined for the six profiles with a 310 μs pulse width during the main stimulation time. The force values were converted into torque by measuring the hip to knee and knee to ankle distances. A Mixed Signal Oscilloscope (Agilent 546222D) and current probe (Philips PM9355) captured the current waveforms during each experiment (set at 30 mA peak).

Six different stimulation profiles were used: -

1) <u>40 Hz Constant frequency</u> (ODSFIII)
2) <u>20 Hz Constant frequency</u>
3) <u>Doublets</u> Double pulses with constant inter-pulse duration. Stimulation is applied at 20 Hz and hence the same number of pulses per second are applied as for the 40 Hz constant frequency (profile 1)
4) <u>Doublets at start and end, 20 Hz in between</u> Modulation of the muscle recruitment
5) <u>Optimum series</u> The same number of pulses per second are applied as for profile 1 (40 Hz) but the inter-pulse intervals are in bursts of 5, 15, 25, 30, 35 and 40 ms over 150 ms (pulses applied at 0, 5, 20, 45, 75, 110 ms). This optimal sequence is based on the tests by Thomas et al [2]. There is evidence that variable spaced pulses can slow muscle fatigue for the same amount of stimulation as constant inter-pulses
6) <u>Optimum series at the beginning and end with 20 Hz in between</u> Optimum series of pulses are applied where the force response in the tibialis anterior would naturally be greatest (similar to profile 4). During these series the simulation is maintained at a constant frequency of 20 Hz

3 Results

Figures 2 to 7 show the torque waveforms for each of the six profiles from one male subject with a normal neurology. The broken horizontal line shows the torque averaged over three waveforms of profile 1.

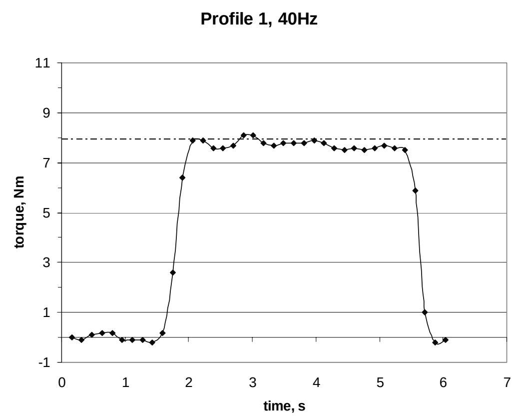


Figure 2

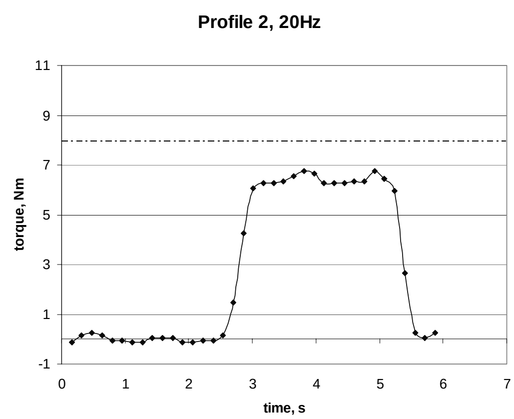


Figure 3

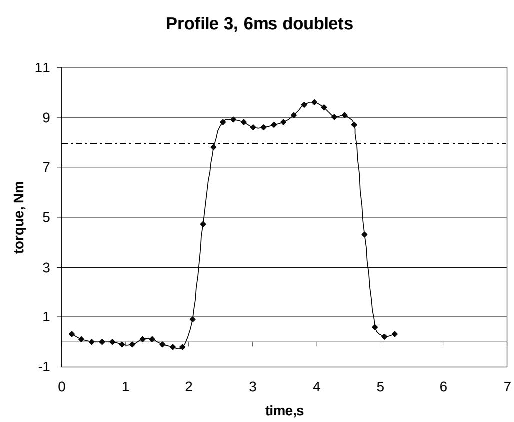
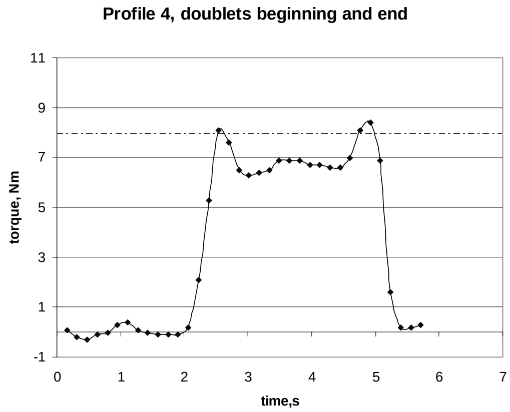


Figure 4



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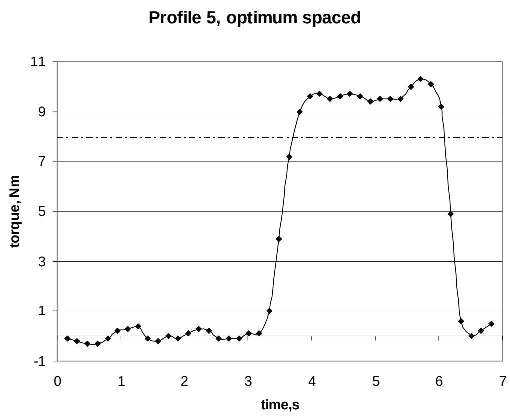

Figure 6

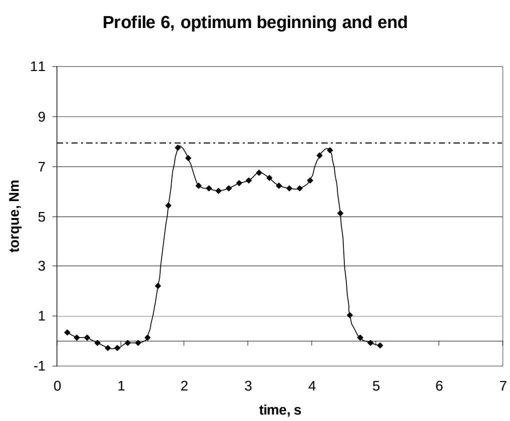

Figure 7

The effect of varying the inter-pulse duration for profile 3 is shown in figures 8 and 9. The averaged values are essentially independent of the duration over the range in this study (figure 9).

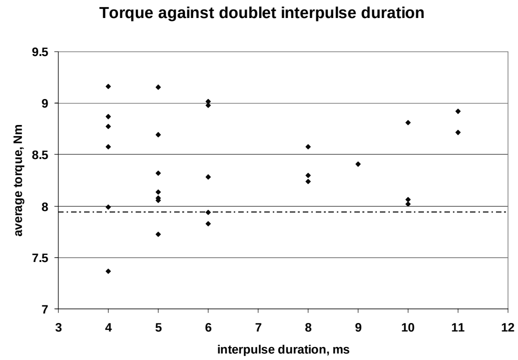

Figure 8

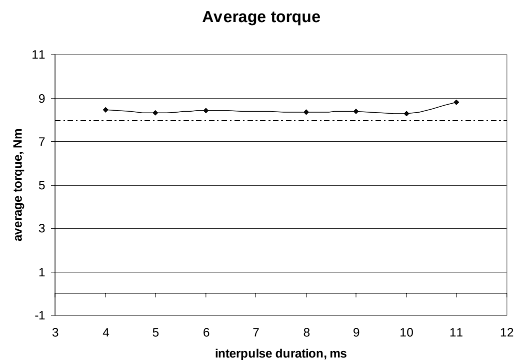

Figure 9

4 Discussion and Conclusions

Doublets enhance the dorsiflexor torque over a range of inter-pulse times (figure 8). The results indicate that including doublets or an optimum series increases the dorsiflexor torque above that produced during regular stimulation (figures 5 and 7). Use of non-repetitive pulses could lead to waveform shaping.

If future experiments are successful in other normal subjects, the test procedure will be repeated with three current ODFSIII users to see if the same results are produced in the target user group.

References

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