Influence of tendon slack on electromechanical delay in the human medial gastrocnemius in vivo

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Muraoka Tetsuro, Tadashi Muramatsu, Tetsuo Fukunaga, and Hiroaki Kanehisa. Influence of tendon slack on electromechanical delay in the human medial gastrocnemius in vivo. J Appl Physiol 96: 540–544, 2004. First published October 3, 2003; 10.1152/japplphysiol.01015.2002.—The purpose of this study was to clarify the influence of muscle-tendon complex stretch on electromechanical delay (EMD) in terms of the extent of tendon slack in the human medial gastrocnemius (MG). EMD and MG tendon length were measured at each of five ankle joint angles (−30°, −20°, −10°, 0°, and 5°; positive values for dorsiflexion) using percutaneous electrical stimulation and ultrasonography, respectively. The extent of MG tendon slack was calculated as MG tendon length shortening, standardized with MG tendon slack length obtained at the joint angle (−16° ± 5°) where the passive ankle joint torque was zero. EMD at −30° (19.2 ± 2.2 ms) and −20° (17.2 ± 1.3 ms) was significantly greater than that at −10° (16.0 ± 2.3 ms), 0° (15.0 ± 1.4 ms), and 5° (14.8 ± 1.4 ms), and at 0 and 5°, respectively. The relative EMD, normalized with the maximal EMD for each subject, decreased dependent on the extent of decrease in MG tendon slack. There were no significant differences in EMD among the joint angles (−10, 0, and 5°) where MG tendon slack was taken up. These results suggest that the extent of tendon slack is an important factor for determining EMD.

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Japan) with their left foot tightly secured by two straps to the dynamometer’s footplate. While the subject maintained completely relaxed leg muscles, the ankle was passively moved cyclically at 5°/s within the range of motion between −35 and 10° (0° was the neutral anatomic position where the sole of the foot was at 90° to the tibia, with positive values for dorsiflexion). Data on joint torque, joint angle, and ultrasonography were collected in the second or later cycle. The test was repeated twice, and the average value obtained was used.

In test B, the subjects lay prone on the dynamometer with their left foot tightly secured to the footplate. While the subject relaxed his leg muscles completely at five different ankle joint angles (5, 0, −10, −20, and −30°), percutaneous electrical stimulation was applied to MG (Fig. 1). In our preliminary experiment on four subjects, we confirmed that there was no significant electrical activity in triceps surae muscles or tibialis anterior muscle within the range of ankle joint angle used in this study (5 to −30°). Surface EMGs from the muscle belly of MG, lateral gastrocnemius, and soleus were recorded by using bipolar surface Ag-AgCl electrodes (5 mm in diameter) with an interelectrode distance of 20 mm. The ground electrode was placed over the lateral malleolus. The electrodes were connected to a pre-amplifier (model 1272, San-ei, Tokyo, Japan; input impedance >200 MΩ, common mode rejection ratio >60 dB) and a differential amplifier having a bandwidth of 5 Hz to 1 kHz (model 1253A, NEC Medical Systems, Tokyo, Japan) to avoid electrical and mechanical noise. EMG data were stored on a personal computer (model Powerbook G3, Apple, Cupertino, CA) via a 16-bit analog-to-digital converter (PowerLab 16/s, ADInstruments, Castle Hill, Australia) at 2 kHz. The pushing force applied to the footplate was measured with a force transducer (type FP/100k, Shinkoh, Tokyo, Japan), which made contact with the ball of the foot. The test was repeated three times at each joint angle. EMD was calculated for each trial, and the average value for each ankle joint angle was used for analysis.

MG contraction was elicited by means of percutaneous electrical stimulation, which lasted for 1,000 ms, through two electrodes (diameter = 20 mm) placed on central and proximal portions of MG. Percutaneous electrical stimulation was used because it can activate only a target muscle (12–14, 21). MG was stimulated with a 100-μs square-wave pulse at 100 Hz by using a stimulator (model SEN-3301, Nihon Koden, Tokyo, Japan) in series with a modified isolator (model SS-1963, Nihon Koden) (13). The stimulation voltage, set as the maximal tolerable voltage determined at the ankle joint angles of −30°, was maintained at all joint angles. Our additional experiment on two subjects showed no significant surface EMG activity in soleus or lateral gastrocnemius muscles by twitch stimulation on MG with a 100-μs square-wave pulse at the same or greater stimulation voltage used in test B (Fig. 1). Surface EMGs were recorded with the acquisition system used for collecting surface EMGs from triceps surae muscles or fibularis longus muscle within the range of ankle joint angle used in this study (5 to −30°).

Fig. 1. A typical raw trace of the electromyograms (EMG) from medial gastrocnemius (MG; A), lateral gastrocnemius (B), and soleus (C) while twitch stimulation is imposed on MG with a 100-μs square-wave pulse at the same or greater stimulation voltage used in test B, which determines electromechanical delay (EMD). The evoked action potential, which appears after the temporal high-voltage pulse, is significant only in MG.

Fig. 2. A typical raw trace of the force signals is shown. Horizontal dashed line, threshold for determining the onset of the pushing force measured at the ball of the foot. Start point of the percutaneous electrical stimulation, 0 ms. Arrow, EMD.

Fig. 3. Typical ultrasonographs over the myotendinous junction region of the MG at plantar flexed position (~30°; top) and dorsiflexed position (5°; bottom). Arrows, myotendinous junction of the MG.
length obtained at the joint angle where the passive joint torque was zero. Thus negative values of the strain represented the extent of tendon slack.

Descriptive data are presented as means ± SD. The effects of the ankle joint angle on EMD and on the threshold level of the onset of the pushing force were examined by using a one-way ANOVA. When the angle-related effect was significant, Tukey’s post hoc analysis was used to determine significant differences between mean values. To test the significance of the relation between the relative EMD and the strain of MG tendon, we calculated Pearson’s correlation coefficient. Statistical significance was set at a level of $P < 0.05$. The reproducibility of the ultrasound and EMD measurements was evaluated on the basis of a coefficient of variation (10, 13, 14–16, 28).

**RESULTS**

The mean CV values of the displacement of P1 between two motions and EMD among three measurements were 3 and 10%, respectively. These values were within the range of previously reported data (10, 13–16, 28).

The threshold level of the onset of the pushing force at 5, 0, −10, −20, and −30° was 0.07 ± 0.01, 0.06 ± 0.01, 0.06 ± 0.02, 0.02 ± 0.02, and 0.07 ± 0.04 N, respectively. There was no significant difference in the threshold level of the onset of the pushing force among different joint angles. EMD averaged between 14.8 ± 1.4 and 19.2 ± 2.2 ms (Fig. 4, A and B). EMD decreased while the ankle joint dorsiflexed (Fig. 4A). EMD at −30° was significantly larger by 4.4 ± 2.2 ms than that at 5°. EMD at −30° was significantly larger than that at −10°, 0°, and 5°. EMD at −20° was significantly larger than that at 0 and 5°. There were no significant differences in EMD among the joint angles of −10, 0, and 5° in the slack tendon region (MG tendon strain < 0%) (Fig. 5). However, in the tight tendon region (MG tendon strain > 0%), there was no significant
correlation between the relative EMD and MG tendon strain (Fig. 5).

DISCUSSION

The present results demonstrate that EMD decreases while MTC length increases until tendon slack is taken up and that EMD obtained in a stretched MTC with a tight tendon remains constant while MTC length increases. These findings support the two hypotheses set at the start of this study.

Before interpreting the results for EMD, a mention should be made of the methodology used to determine the onset of the threshold of the force signal (pushing force) and the tendon slack length. The threshold of the force signal in EMD analysis is one of the factors that affect its duration. A high threshold may result in a greater delay time, which is an artifact. The time for elongating SEC is affected by the rate of force development, which is affected by MTC length (11). Thus a high threshold would make it difficult to show the influence of SEC slack on EMD. In the present study, the threshold was defined as the first point to rise above the 99% confidence interval of baseline for 20 ms and was 0.07 N on the average, which was quite low (Fig. 2). Therefore, it seems reasonable to consider that the major portion of the time for elongating SEC in our EMD measurement was the time used to take up SEC slack. In the present study, it was assumed that the tendon slack length was obtained at the joint angle where the passive ankle joint torque was zero. The passive joint torque is affected by all soft tissues surrounding the joint. Riener and Edrich (20) examined the relationship between ankle joint angle and passive ankle joint torque with the knee flexion angles of 0 and 60°, and they showed that 1) the passive ankle joint torque with the knee flexion angle of 0° was zero around the ankle joint angle of −18°, which was similar to our results, and 2) when the ankle joint angle was below about −18° to −22°, the passive ankle joint torque was not significantly affected by knee joint angle (gastrocnemius length). These results indicated that gastrocnemius was almost slack when the passive ankle joint torque with the knee flexion angle of 0° was below zero. Therefore, it seems reasonable to calculate the tendon strain using the tendon length at the neutral angle (angle at which passive torque is zero) as the reference length, although the tendon strain and the extent of tendon slack might be slightly underestimated and overestimated, respectively.

In the present study, EMD averaged between 14.8 ± 1.4 and 19.2 ± 2.2 ms (Fig. 4, A and B). These values were at the lower end of the generally reported values of 7–122.9 ms (3, 5, 6, 11, 15, 18, 19, 26, 29) but were comparable with those (7–18.77 ms) reported by Moritani et al. (15), Muro and Nagata (18), and Zhou et al. (29), who used electrical stimulations. The EMD measured using electrical stimulation is shorter than the EMD measured in voluntary contractions (26, 29). The shorter EMD measured in the present study could be attributed to synchronized motor unit activity and to reversed recruitment order (23), as discussed by Zhou et al.

The present results showed that EMD was independent of joint angles where tendon slack was fully taken up by MTC stretch (Fig. 4B and 5), although there was a trend toward a decrease in EMD with an increase in ankle joint angle (Fig. 4A), and that EMD decreased while the extent of tendon slack decreased by MTC stretch (Fig. 5). MTC stretch therefore may not influence EMD when EMD is measured within the joint angle range where tendon slack is taken up. The knee joint angle ranges adopted by Granata et al. (6) and Vos et al. (26), who reported that MTC stretch did not influence EMD in the quadriceps muscles, were 45–90 and 90–130° (0° was full knee-extended position), respectively. The tendon slack of the quadriceps muscles was eliminated in those joint angle ranges because a passive knee extension torque was observed (20) and the vastus lateralis muscle was on the descending limb of its force-length relation (9). Therefore, the present results were in line with their findings about the influence of MTC stretch on EMD. The present results also suggest that MTC stretch may influence EMD when EMD is measured within the range of joint angles including the range where the tendon is slack. The hip joint angle range adopted by Laine Santa Maria (11), who reported that MTC stretch influenced EMD, was 0–90° (0° was full hip-extended position) with the knee flexed by 90°. The tendon was slack in a part or all of this joint angle range, judging from the passive joint torque-joint angle relation (20). Thus the present results were also in line with the finding from Laine Santa Maria. Tendon slack may partly explain the inconsistency in findings about the effect of MTC stretch on EMD in previous studies (6, 11, 26).

EMD at −30° was greater by 3.2 ms (20%) than that at −10° where tendon slack was fully taken up (Fig. 4B). The influence of MG tendon slack on EMD in the normal range of the ankle joint might be considered small when EMG data are analyzed in relation to kinetic and/or kinematic data. As described earlier, however, the use of electrical stimulation resulted in shorter EMD times compared with the times obtained via voluntary contractions (29), judging from the previous findings that EMD measured during voluntary contractions (3, 5, 6, 11, 19, 26) is about four times greater than that measured using electrical stimulation (15, 18, 29). It can therefore be speculated that the time to take up tendon slack in voluntary contractions is greater than that in the contractions elicited by electrical stimulation. Consequently, the time to take up tendon slack may be important in the analysis of EMG data in relation to kinetic and kinematic data.

In summary, the present results demonstrate that EMD in the human MG is independent of joint angles where MG tendon slack is fully taken up by MTC stretch and that EMD obtained at a MTC length with a slack tendon is greater than that at a MTC length with a tight tendon. Therefore, it is suggested that the extent of tendon slack is an important factor for determining the force transmission capability of tendons (i.e., the time required to transmit contraction forces to bones).

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REFERENCES


