PRELIMINARY ANALYSIS OF SLEDGE REACTION FORCES DURING CYCLICAL LOADING OF THE TRICEPS SURAE

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To date, no methodology exists that can measure Achilles tendon stiffness in a controlled dynamic situation while simultaneously investigating tendon and joint stiffness interactions. Stiffness refers to the ratio between force and elongation, and the aim of this preliminary study was to establish an analysis protocol for sledge reaction forces during cyclical loading of the triceps surae. Results indicated the magnitude of forces was just under 50% of body weight, which was expected. Removal of the first two trials from analysis reduced standard deviation and 95% confidence interval of plantarflexor force, contact time and contact time-flight time ratio, suggesting this protocol is suitable to ensure data gathered is repeatable and consistent. Future work using inverse dynamics and ultrasound shall provide information on Achilles tendon loading and stiffness.

KEYWORDS: biomechanics, Achilles tendon, tendon mechanics, ultrasound, stiffness

INTRODUCTION: Stiffness refers to the ratio between force and elongation. The importance of leg and joint stiffness in sporting performance has previously been established (Hobara et al., 2010; Kuitunen et al., 2002), and Achilles tendon stiffness has been shown to be related to both injury and performance (Kubo et al., 2000; G. A. Lichtwark & Wilson, 2008; Mahieu et al., 2006). Tendon stiffness can be measured indirectly using modelling (Kafka et al., 1995), electromechanical delay (Winter & Brookes, 1991) and the quick release method (Rabita et al., 2008) or using invasive and non-invasive direct methods (Arampatzis et al., 2007; G.A. Lichtwark & Wilson, 2005). Isometric dynamometry and ultrasound are often used to determine tendon force and elongation, but the relationship between tendon mechanics during isometric and concentric contractions is questionable. Studies of concentric conditions typically have little control over subject movement during the activity itself which may also affect loading rates and patterns. Previous work relating tendon stiffness during isometric contractions to joint stiffness in dynamic situations is also potentially erroneous (Kubo *et al.*, 2007).

Force sledges have previously been used to investigate differences in stretch-shortening cycle (SSC) function (Harrison et al., 2004) and to perform fatiguing SSC exercise (Ishikawa *et al.*, 2006). Finni *et al.* (2000) used a force sledge and ultrasound to investigate *in vivo* muscle mechanics, with patellar tendon forces measured using invasive implanted optic fibres. Tendon stiffness or kinematic data was not reported however. This suggests a need to develop a non-invasive, controlled and direct method of determining *in vivo* tendon mechanics during dynamic activities that can also allow study of the tendon-joint stiffness relationship.

To date, there is no normative data for the magnitude of plantarflexor force during rhythmical cyclical loading using a novel adaptation to a force sledge. This SSC activity is thought to be similar to that of hopping or running. The consistency and repeatability of these forces are also unknown. This preliminary study investigated the reaction forces during repeated impacts using a force sledge and aimed to establish a protocol for data analysis. This protocol will then be used in further studies using the sledge to investigate tendon stiffness and the tendon-joint stiffness interaction during dynamic activities.

METHODS: Data collection: Following university ethics committee approval, seven trained males (mean ±SD: age: 22.1 ±1.4 years; height: 183.9 ±6.0 cm; mass: 87.0 ±7.5 kg) volunteered as subjects for this study. None had suffered an ankle injury in the 3 months prior to testing or had ever had surgery of the lower limb. Subjects completed a Physical Activity Readiness Questionnaire and signed an informed consent prior to participation.

19 mm passive reflective markers were placed on both sides of the plate, and three motion analysis cameras (Motion Analysis Corporation, Santa Rosa, CA, USA) sampling at 500 Hz were set up on each side of the sledge to measure plate displacement. The test area was calibrated using an L-frame and 500 mm wand with calibration accepted when average wand length was 500 ± 0.04 mm (deviation <0.68 mm) and average 3D residual was <0.33 mm (deviation <0.46 mm).

Following familiarisation, trials began with a wooden plate resting on the foot. This plate was at 90° to the sledge rails and was free to move up and down. Subjects lay on a solid plinth secured to the base of the force sledge. The subject was secured so force generated came from the triceps surae. The thigh of the test limb was secured to a wooden support using two Velcro straps and a harness was placed over the anterior superior iliac spines to minimise hip movement.

Based on previous work in this lab, the plate was lifted to a position 30 cm above the foot while the sole of the foot was at 90° to the tibia. The plate was released following a 3-2-1 countdown. Subjects were instructed to contact the plate and push it back up the rails as rhythmically and continuously as possible while minimising contact time. They were also instructed to aim for a marked square on the plate. All subjects were given the same instructions due to previously reported changes in stiffness modulation with different instructions. Each trial consisted of 5 impacts.

Data analysis: All data was exported to Microsoft Excel (Microsoft Inc., USA) for analysis. Marker displacement was filtered using a 10 Hz low-pass reverse pass Butterworth filter. Marker acceleration was calculated as the second derivative of filtered displacement values. Force was calculated using Newton's second law and accounted for sledge slope of 30°.

Statistical analysis: Mean force, contact time (CT) and contact time: flight time ratio (CT: FT ratio) was calculated for each trial to determine the range of forces generated by the plantarflexors. Standard deviations (SD) and 95% confidence intervals (95%CI) were calculated to determine the variability of these measures. These statistics were calculated using all impacts in the trial, and then repeated with the first and second impacts removed to determine which trials should remain in the analysis protocol.

Table 1

Mean, standard deviation and 95% confidence interval for plantarflexor force

	All five impacts	**First impact removed**	**Second impact removed**
Mean (N)	459.8	462.5	471.7
Standard deviation (N)	35.88	29.05	21.20
Standard deviation (% of mean)	8.24	6.63	4.74
95% confidence interval (N)	70.3	56.9	38.1
95% confidence interval (% of mean)	16.15	13.00	8.47

Table 2

Mean, standard deviation and 95% confidence interval for plate contact time

	All five impacts	**First impact removed**	**Second impact removed**
Mean (ms)	14	14	13
Standard deviation (ms)	1	1	1
Standard deviation (% of mean)	7.6	7.4	6.2
95% confidence interval (ms)	2	2	2
95% confidence interval (% of mean)	14.9	14.5	11.2

Table 3

Mean, standard deviation and 95% confidence intervals for plate contact time: flight time ratio

	All five impacts	First impact removed	Second impact removed
Mean	0.23	0.23	0.23
Standard deviation	0.03	0.02	0.01
Standard deviation (% of mean)	10.08	6.58	5.38
95% confidence interval	0.05	0.03	0.02
95% confidence interval (% of mean)	19.75	12.90	10.54

RESULTS AND DISCUSSION: Results are as shown in tables 1 to 3. Forces exerted were just under 50% body weight, which is approximately one quarter that which would be expected during a one-legged vertical jump (Cordova & Armstrong, 1996). This is probably because the lower limbs are not loaded by body weight during this protocol and force is solely produced by the plantarflexor muscle group. Mean plantarflexor force increased with removal of the initial impacts and SD reduced from 8.24% to 7.47% of the mean. The 95%CI around the mean also reduced, suggesting more repeatable, consistent impacts. Reduced SD and 95%CI of CT and CT:FT ratio with removal of the earlier impacts further suggests this is necessary to ensure repeatable, consistent data.
These changes are most likely due to different muscle activation patterns during the first, second and subsequent impacts. Muscle activation has previously been shown to vary dependent on task demands (Wakeling, 2004). The subject could see the plate at all times, but it is likely that during the initial impacts regardless of the amount of familiarisation subjects had, their initial level of pre-activation was different to that of subsequent impacts. During the second and subsequent trials, subjects were getting used to the loading on the plantarflexors hence the muscle activation response was optimised to that condition. This form of analysis is similar to that of Lichtwark and Wilson (2005) who only analysed ~5 of the 15-20 hops each subject was asked to perform.

CONCLUSION: The results of this study suggest that a force sledge can be used to determine plantarflexor force and that a number of impacts are required for data analysis as there is a different muscle response during the first two impacts. Further study including use of inverse dynamics and ultrasound will allow for calculation of in vivo Achilles tendon force, elongation and hence stiffness. Use of this methodology can then be used to investigate the contribution of tendon stiffness to joint stiffness and hence performance, as well as allow for investigation of Achilles tendon injuries in a controlled environment.

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