Relationship Between Balance Recovery From a Forward Fall and Lower-Limb Rate of Torque Development

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Relationship between balance recovery from a forward fall and lower-limb rate of torque development

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Abstract

The authors examined the relationship between the maximum recoverable lean angle via the tether-release method with early- or late-phase rate of torque development (RTD) and maximum torque of lower-limb muscle groups in 56 young healthy adults. Maximal isometric torque and RTD at the hip, knee, and ankle were recorded. The RTD at 50-ms intervals up to 250 ms from force onset was calculated. The results of a stepwise multiple regression analysis, early RTD for hip flexion, and knee flexion were chosen as predictive variables for the maximum recoverable lean angle. The present study suggests that some of the early RTD in the lower limb muscles, but not the maximum isometric torque, can predict the maximum recoverable lean angle.

Key words: Balance recovery; Rate of torque development; Maximum isometric torque; Maximum recoverable lean angle

1. Introduction

The ability to step forward rapidly with the lower limb plays an important role in preventing a fall after forward loss of balance (Van Dieën, Pijnappels, & Bobbert, 2005). In the tether-release method, which is used to investigate step recovery in fall avoidance, the subject is placed in a forward inclined position with hips pulled backwards (Hsiao-Wecksler, 2008). Individuals who can recover their balance in a single step from a maximum initial forward leaning position, known as the maximum recoverable lean angle, have a better ability to recover balance (Thelen, Wojcik, Schultz, Ashton-Miller, & Alexander, 1997). Several previous studies using the tether-release method revealed that older adults have less maximum recoverable lean angle compared to young individuals (Hsiao-Wecksler & Robinovitch, 2007; Madigan & Lloyd, 2005; Thelen et al., 1997; Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 1999). Additionally, older adults are more likely to use multiple steps to recover balance as the initial forward lean angle increases (Carty et al., 2015; Carty, Barrett, et al., 2012; Carty, Cronin, Lichtwark, Mills, & Barrett, 2012). In older adults, the use of multiple steps to recover balance during tether-release experiments is a predictor of future fall events (Carty et al., 2015).

Several studies have attempted to predict the maximum recoverable lean angle or magnitude using the maximum joint torque of the lower limb. In a study of young and older adults, isometric torques of hip flexion and ankle plantarflexion were not good predictors of maximum recoverable lean angle (Wojcik, Thelen, Schultz, Ashton-Miller, & Alexander, 2001). In contrast, other studies showed that the maximal isometric joint torques of ankle plantarflexion and knee extension could predict the margin of stability in young and older adults (Karamanidis, Arampatzis, & Mademli, 2008). Furthermore, ankle dorsiflexion torque is also a weak predictor of balance recovery in older adults (Grabiner, Owings, & Pavol, 2005). In a recent study (Graham, Carty, Lloyd, & Barrett, 2015) amongst community-dwelling older adults, which used a stepwise multiple regression to analyze maximal recoverable lean magnitude as the independent variable, some joint moments and powers in the stepping leg during balance recovery were extracted as explanatory variables, whereas isometric joint torques were not. These studies have all measured maximum joint torques of the lower limb using an isokinetic dynamometer. Thus, it is not clear if maximum joint torques is a good predictor of maximum recoverable lean angle. Balance recovery requires the timely generation of appropriate joint moment and muscle power to step forward quickly (Aragão, Karamanidis, Vaz, & Arampatzis, 2011; Arampatzis, Peper, & Bierbaum, 2011; Madigan, 2006); thus, apart from muscle strength, explosive force was also thought to be necessary for rapid stepping.

The relationship between the maximum recoverable lean angle and maximum isometric torque of the lower limb has been frequently investigated, whereas the rate of torque development (RTD), which is the rate at which torque production occurs, has not been investigated. The characteristics of RTD are inconsistent during contraction. A relatively early-phase RTD within the first 100 ms of a rapid contraction shows great variability between different individuals (Folland, Buckthorpe, & Hannah, 2014), while a late-phase RTD of a longer duration (100-250 ms) has a strong correlation with maximum muscle strength (Andersen & Aagaard, 2006). The early phase of RTD is related to neuronal factors like individual motor unit discharge rate. Since this is the chief force-generating capacity in an explosive situation, it likely plays an important role in fall avoidance (Maffiuletti et al., 2016).

The aim of this pilot study was to investigate the correlation between the maximum recoverable lean angle, using the tether-release method with maximum torque, and RTD in each phase and each joint of the stepping limb. Fall avoidance relies on production of adequate voluntary muscle strength in a short period of time. In addition, achieving balance recovery from a larger initial lean angle requires faster joint velocity (Madigan & Lloyd, 2005) and greater muscular activity (Thelen et al., 2000). We hypothesized that early RTD will be a better predictor of maximum recoverable lean angle than late RTD or maximum isometric torque of the lower limbs.

2. Methods

2. 1. Participants

The participants comprised 56 untrained healthy young adults (28 men; mean age, 21.0 ± 0.8 years; height, 1.70 ± 0.05; weight, 62.1 ± 7.2 kg; 28 women; mean age, 21.1 ± 0.8 years; height, 1.55 ± 0.06; weight, 49.0 ± 4.7 kg). People with orthopedic disorders that would impede fall-avoidance stepping performance were excluded. Furthermore, targeted participants were free of upper and lower limb pain and discomfort. G*Power (ver. 3.1.9.2) was used to determine the sample size. To calculate the sample size of a multiple regression analysis, we used Cohen's f^2 for effect size, set at 0.35 (representing a large effect) and at an alpha level of 0.05 and power of 0.80. The number of predictors was set at 30, as RTD consists of 6 joint movements and 5 time points. Based on the above assumptions, a minimum of 36 participants were required for this study. The study was approved by the Seijoh University Ethics Committee (Approval Number: 16PT07) and informed consent was obtained from all participants who received sufficient explanation about the research objectives and methods.

2. 2. Experimental procedures

2. 2. 1. Measurements of maximum recoverable lean angle

Participants were fitted with a harness (Full harness EHC-9A, Sanko, Inc., Japan) and a tether was attached at the posterior lumbar L1-L2 level. The tether release switch, a customized car seatbelt buckle, was fixed to a metal strut that permitted height adjustment behind the subject. While tethered, with arms folded the chest, the participants adopted a forward inclined ready-position with legs placed horizontally and shoulder-width apart. A Chapman dominant leg test (Chapman, Chapman, & Allen, 1987) was performed, and the leg that used for stepping during the fall was defined as the dominant or stepping leg. Participants were instructed in advance to use their dominant leg during the stepping movement. Reflective markers were attached to the acromion and lateral malleolus on the stepping-leg side. An optical, high-speed camera synchronized to a personal computer was installed at a position 2 m lateral to the stepping-leg side. The camera frame rate was 240 fps. Participants were instructed to keep their back straight in the forward inclined position prior to tether release. The forward inclination angle (Hsiao-Wecksler & Robinovitch,

2007) between the axis perpendicular to the floor and the line connecting the acromion and lateral malleolus markers on the stepping-leg side was derived using a free imaging analysis software (ImageJ, version 1.44). For safety, a cushioned mat was placed 2 m in front of the subject.

Participants were instructed to quickly move their stepping leg forward at the instant the tether was released and limit this movement to 1 step. Forward inclination angle was increased by 5º increments starting from 15º until single-step balance recovery was no longer possible, or a portion of their body touched the cushioned mat in front of them. After failing twice in the single step balance recovery, the forward inclination angle was then reduced by 2º increments until balance recovery was again successful twice, which was defined as the maximum recoverable lean angle.

2. 2. 2. Torque measurements

A hand-held dynamometry (HHD) (Mobie MT-100, SAKAImed, Japan) and pull sensor (MT-150, SAKAImed, Japan) were used for torque measurements of flexion and extension in the hip, knee, and ankle (Fig. 1). The device senses and measures force by pulling a distortion gauge, and joint torque can be measured with a fixation to the non-elastic belt (Suzuki, 2015). Hence, the HHD with external fixation was used in this study so that the examiner is not required to hold the HHD. The lower limb of the stepping side during balance recovery tasks (i.e., dominant leg) was measured. The positions of the joints for each of the force measurements by the HHD are shown in Fig. 1. Limb position and the belt with pull sensor installation locations (i.e., points of resistance) were based on the methods of Thorborg et al. (2013), Koblbauer et al. (2011), and Moraux et al. (2013) for measurements involving the hip, knee, and ankle dorsiflexion, respectively. The participants were seated on a plinth adjustable to their height in an upright position and their hips and knees were positioned at approximate right angles. The joint angles were measured with a goniometer based on the body landmark (e.g., line connecting the greater trochanter, knee joint center, lateral malleolus) in the testing positions. The belt with pull sensor was positioned distally on the thigh, distally on the anterior aspect of the tibia, distally on the posterior calf complex, and on top of the foot at the level of the metatarsal, for hip flexion, knee extension, knee flexion, and ankle dorsiflexion, respectively. For hip extension, the belt was positioned at the posterior calf-complex with participants in a prone position. Ankle plantarflexion torque had to be measured with knee extension, because the stepping reaction from a forward fall required a push-off in the knee extension position. Specifically, for ankle plantarflexion, the participants were positioned directly on an isokinetic joint torque measuring device in a long sitting position with hips flexed at 70°, knees extended at 0°, trunk and thighs fixed, and the belt with pull sensor installed between the planta and the ankle plate. To ensure muscular contraction without joint movement, the belt with the pull sensor was tautened to keep the limb in the torque measurements position. The length of the lever arm, which spanned the distance between the center of the joint and the point of effort, i.e., the location of the belt with pull sensor, was recorded for each subject in all measurements. A previous study has reported that the rate of force development measured using the HHD has a high reliability (Mentiplay et al., 2015).

+++++ Include Figure 1 here +++++

Participants performed a sufficient warm-up and three rounds of practice trials with moderate effort before measurement. To calculate the isometric maximum joint torque and RTD of these joint movements, participants were instructed to quickly exert maximum isometric joint torque when a signal was given by the examiner. Strong verbal encouragement was provided during each joint torque measurements to promote maximal effort. Force values were continuously recorded at a sampling rate of 1.5 kHz using the Myoresearch version 2.1 (Noraxon USA, Inc., Scottsdale, AZ). Each joint movement was successively measured three times.

2. 3. Data analyses

Force data were band-pass filtered at 20–500 Hz with second-order Butterworth characteristics, and multiplied by lever-arm length and divided by subject body weight to derive a normalized torque-time curve. Maximum joint torque was defined as normalized torque-time curve peak values (Nm/kg). The average value of three maximal joint torque was adopted used for the final analyses.

The time of torque onset was defined as the moment when the HHD reading exceeded three standard deviations (SD) below the average value during the 500 ms before force exertion, based on the methods of de Ruiter, Kooistra, Paalman, and de Haan (2004). In addition, onset was visually verified for each subject. The slope of the torque-time curve was calculated (Nm/kg/s) from onset with every 50 ms interval up to 250 ms, named $RTD_{0\text{-}50}$, $RTD_{0\text{-}100}$, $RTD_{0\text{-}150}$, $RTD_{0\text{-}200}$, and $RTD_{0\text{-}250}$. The average value of three RTDs for each time point was used for the final analyses.

2. 4. Statistical processing

The normality of all data was examined with the Shapiro-Wilk test. All data, including the maximum joint torques, each time point of RTD, and the maximum recoverable lean angle, were normally distributed. The intra-rater reliability of the maximum joint torque and RTD at each time point among three measurements was estimated using intra-class correlation coefficients (ICC). Pearson's product moment correlations assessed the relationships between maximum recoverable lean angle and each time point on RTD-dependent variables. The Pearson product moment correlations were presented for all RTD at each time point and maximum joint torque, and multicollinearity was verified prior to multiple regression analyses. If the correlation coefficient between the two RTD in the same joint movement exceeded 0.80, the later RTD was excluded from the explanatory variables. Multiple stepwise regression analyses were performed using maximum recoverable lean angle as the independent variable and lower-limb maximum joint torque and each time point on RTD as explanatory variables. The statistical significance threshold was set at 5%.

3. Results

The ICCs for the maximum joint torque ranged from 0.90 to 0.96 for each targeted joint movement. Additionally, the ICCs for the RTD at each time point and among targeted joint movements are provided in Table 1. Although early RTD (≥100 ms) for some joint movements exhibited a lower value compared with late RTD, these ICC results indicated substantial to almost perfect reliability (Landis & Koch, 1977).

+++++ Include Table 1 here +++++

The average ± SD for maximum recoverable lean angle was 32.4 ± 5.1° in all participants. Maximum joint torque and RTD data at each time point are shown in Table 2. A significant positive correlation was observed for the maximum recoverable lean angle with hip flexion ($r = 0.561$, Cohen's $f^2 = 0.46$), hip extension ($r = 0.301$, Cohen's $f^2 = 0.10$), knee flexion ($r = 0.341$, Cohen's $f^2 = 0.13$), ankle plantarflexion ($r = 0.334$, Cohen's $f^2 = 0.13$), and ankle dorsiflexion ($r = 0.538$, Cohen's $f^2 = 0.41$) by maximum joint torque. As shown in Table 3, a significant positive correlation was observed for RTD of each time point on several of these. All the hip flexion RTDs at each time point showed a significant relationship, while significant relationships were not found for all the knee extension RTDs at each time point.

+++++ Include Table 2 and Table 3 here +++++

The $RTD_{0\text{-}200}$, $RTD_{0\text{-}250}$, and maximal joint torque in all joint movements had a correlation coefficient of more than 0.80. This means that the maximum torque and RTD_{0-200} and RTD_{0-250} were strongly correlated. Therefore, RTD_{0-200} and RTD_{0-250} were excluded from the explanatory variables to avoid multicollinearity. Instead, the maximum joint torque was included. Multiple stepwise regression analysis showed that hip flexion RTD_{0-50}, knee flexion RTD_{0-100}, and hip flexion RTD_{0-150} (adjusted R^2= 0.589, F= 27.27, p< 0.001) were the best predictors of maximum recoverable lean angle (Table 4).

+++++ Include Table 4 here +++++

4. Discussion

To the best of our knowledge, this is the first study to examine the relationship between the maximum recoverable lean angle created by the tether-release method and RTD for the lower limb. Our results support our hypothesis that early-phase RTD predicts the maximum recoverable lean angle better than maximum isometric torque. Maximum recoverable lean angle was correlated with maximum isometric torque and RTD for some joint movements, but not knee extension in the single regression analysis. A stepwise multiple regression analysis involving RTD less than 200 ms and maximal joint torque showed that hip flexion RTD_{0-50} and RTD_{0-150} as well as knee flexion RTD_{0-100} were predictors of maximum recoverable lean angle, as opposed to maximum isometric torque. Additionally, the standard partial regression coefficient displayed a stronger effect in the

RTD_{0-50} and RTD_{0-100} of the hip and knee flexion than the RTD_{0-150} of hip flexion.

Single regression analysis showed that the maximum isometric torque, excluding knee extension, significantly correlated with maximum recoverable lean angle. The effect of maximum muscle strength on RTD increases with time from the onset of contraction; particularly as RTD_{0-200} has a strong correlation with maximum muscle strength (Andersen & Aagaard, 2006). This may explain our results indicating that the maximum isometric torque, excluding knee extension, and RTD_{0-200} and RTD_{0-250} in the same joint movement were both significantly correlated with the maximum recoverable lean angle. Although there were significant positive correlations in most of the joint movements, none were chosen as predictors of maximum isometric joint torque in the stepwise multiple regression analysis. Maximum available torques in the stepping leg were not used during the balance recovery from tether-release in younger adults (Graham, Carty, Lloyd, Lichtwark, & Barrett, 2014; Wojcik et al., 2001). Therefore, as an individual's maximum torque level does not directly relate to the balance recovery capacity, isometric maximum joint torque is only at most a moderate predictor of maximum recoverable lean angle.

Reduced postural stability during upright standing in older adults is related to decreased leg extension rate of force development (Izquierdo, Aguado, Gonzalez, López, & Häkkinen, 1999). Decreased production of explosive force might affect the time until neuromuscular response during balance recovery. The muscle reaction time for the stepping limb in tether release was within 80 ms (Thelen et al., 2000). Therefore, early RTD of lower limb joint torque is likely involved in impulsive situations such as fall avoidance. In fact, early RTD, namely the RTD_{0-50} of hip flexion and the RTD_{0-100} of knee flexion, were extracted as predictors of the maximum recoverable lean angle in this study. It has been reported that early RTD is predominantly dependent on muscular activation levels at the onset of the contraction (de Ruiter et al., 2004). Recruiting a larger proportion of the available motor units is required to achieve a large and rapid stepping movement during balance recovery (Cronin, Barrett, Lichtwark, Mills, & Carty, 2013). The lower rate of development for muscle activation has been shown to lead to decreased rate of force generation in the lower leg, resulting in an inadequate recovery response and increased fall risk (Pijnappels, Bobbert, & Van Dieën, 2005). The hip flexion and knee flexion torques, chosen as predictive variables in the current study, work in the early phase during the tether release step, and contribute to forward progression and knee flexion in the stepping limb (Madigan, 2006). The rate of hip flexion moment generation during balance recovery is related to the maximum recoverable lean angle magnitude for tether-release (Arampatzis et al., 2011). Another study also reported that the semitendinosus peak muscular activity contributing to knee flexion was significantly associated with step length during balance recovery (Cronin et al., 2013). The relationships between the balance recovery capacity and the lower limb early RTD in the current study may be indirectly related to the ability to execute large and rapid steps.

In a previous study of lower limb torques measured by an isokinetic dynamometer and a simple linear regression analysis of balance recovery, the margin of stability for joint torques of the ankle plantarflexion and knee extension were predicted as 44% and 35%, respectively (Karamanidis et al., 2008), and ankle dorsiflexion torque predicted maximum recoverable lean angle in older adults at a rate of 30% (Grabiner et al., 2005). Moreover, ankle plantarflexion and hip flexion muscle strength predicted the maximum recoverable lean magnitude at contribution rates of 18% and 19%, respectively (Graham et al., 2015). Although this study of healthy young volunteers differs from the studies that included older adults, the $RTD_{0\text{-}50}$ of hip flexion and $RTD_{0\text{-}100}$ of knee flexion, and the $RTD_{0\text{-}150}$ of hip flexion that were measured by a HHD predicted the maximum recoverable lean angle at a multiple coefficient of determination of 59%. The comprehensive analysis including maximum isometric torque of the lower limbs and RTD in this study demonstrated that maximum recoverable lean angle can be predicted. The relationship between explosive force and maximum recoverable lean angle, including kinematic analysis of older adults, needs to be investigated in the future.

When interpreting the results of the present study, caution is needed regarding the following limitations. First, since the joint angles at peak contraction was not confirmed, participants may have been allowed a slight movement of the joint during the explosive maximum torque measurement, with the exception of ankle plantarflexion. Participants kept the limb position with the HHD belt taut at the position in which maximum torque was produced. This could cause slight muscular activation, which might have affected maximum joint torque. Nevertheless, the RTD at each time point and each joint had moderate to high reproducibility even if there are limitations of the method used for joint torque measurements in the current study. Second, the joint torque measurements used were isometric contractions and do not reflect the joint angular speed pertinent to balance recovery stepping. Third, although there may be a gender difference in magnitude of joint torques used for balance recovery stepping (Wojcik et al., 2001), the regression analysis in the current study included men and women. Lastly, as no kinesiologic or electromyographic analysis of the tether-release method was conducted, it remains unclear how the participants' joint strength contributed, or how muscle co-activations or coordination of contraction timing may have affected balance recovery. Forward balance loss recovery was accomplished by adequate trunk regulation, lower limb moment generation, power, and a long and rapid step (Graham et al., 2015). Accordingly, we agree that predictor variables for maximum recoverable lean angle, including kinematic analysis of tether release stepping must be determined. In particular, it is necessary to clarify the explosive force of lower limbs that contributes to the expansion of the step length from the maximum recoverable lean angle.

5. Conclusion

RTD measurement using the HHD is a predictive factor for maximum recoverable lean angle in the tether-release test. Additionally, hip flexor $RTD_{0\text{-}50}$, $RTD_{0\text{-}150}$, and knee flexor $RTD_{0\text{-}100}$ were related to 59% of the shared variance of maximum recoverable lean angle. The findings of the present study suggest that early-phase RTD for a portion of the lower limb, rather than maximum isometric torque, can predict maximum recoverable lean angle in healthy young adults.

Conflict of interest statement

None of the authors report a conflict of interest.

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Figure 1.

Figure title and caption

Figure 1: Testing positions for force measurements for the dynamometer and belt with pull sensor.

The belt with the pull sensor was fixed to the metal base frame placed on the floor for hip flexion (A), hip extension (E), and ankle dorsiflexion (G). The belt was also externally fixed to the vertical metal bar or a plinth frame for knee flexion (B) or knee extension (F), respectively. For the measurements of ankle plantar flexion force (C), the belt with pull sensor was fixed to the seat frame of the isokinetic joint torque measuring device to be straight along the long axis of the lower leg. The belt with the pull sensor and hand-held dynamometry (D), and the device in action (H).

Table 1

Table 1. Intra-class correlation coefficients for the maximum joint torques and rate of torque development at each time point.

	Peak torque	$RTD_{0\text{-}50}$	$RTD_{0\text{-}100}$	$RTD_{0\text{-}150}$	$RTD_{0\text{-}200}$	$RTD_{0\text{-}250}$
HF	0.90 (0.85-0.94)	0.81 (0.71-0.89)	0.83 (0.73-0.89)	0.86 (0.79-0.92)	0.86 (0.78-0.91)	0.87 (0.79-0.92)
HE	0.95 (0.92-0.97)	0.77 (0.64-0.86)	0.82 (0.73-0.89)	0.83 (0.74-0.90)	0.92 (0.87-0.95)	0.87 (0.80-0.92)
KF	0.96 (0.94-0.98)	0.82 (0.71-0.89)	0.86 (0.78-0.91)	0.83 (0.74-0.90)	0.94 (0.90-0.96)	0.93 (0.89-0.96)
KE	0.91 (0.86-0.94)	0.74 (0.60-0.84)	0.84 (0.75-0.90)	0.83 (0.74-0.90)	0.88 (0.82-0.93)	0.89 (0.83-0.93)
APF	0.90 (0.85-0.94)	0.74 (0.59-0.84)	0.82 (0.72-0.89)	0.81 (0.70-0.88)	0.81 (0.71-0.88)	0.87 (0.79-0.92)
ADF	0.92 (0.87-0.95)	0.73 (0.57-0.83)	0.82 (0.73-0.89)	0.87 (0.80-0.92)	0.93 (0.89-0.96)	0.90 (0.84-0.94)

These data were shown in the ICCs (95% confidence intervals from lower bound to upper bound).

HF, hip flexion; HE, hip extension; KF, knee flexion; KE, knee extension; APF, ankle plantarflexion; ADF, ankle dorsiflexion.

Table 2

Table 2. Mean lower limb maximum joint torques and rate of torque development at each time point (mean ± standard deviation).

	Maximum joint torque	$RTD_{0\text{-}50}$	$RTD_{0\text{-}100}$	$RTD_{0\text{-}150}$	$RTD_{0\text{-}200}$	$RTD_{0\text{-}250}$
HF	1.46 ± 0.29	9.71 ± 4.07	8.92 ± 3.14	5.76 ± 1.67	5.66 ± 1.24	4.53 ± 0.91
HE	1.42 ± 0.38	10.09 ± 4.43	9.16 ± 3.05	5.80 ± 1.65	5.62 ± 1.62	4.38 ± 1.14
KF	1.42 ± 0.41	9.81 ± 4.88	8.75 ± 3.32	5.70 ± 1.84	5.49 ± 1.64	4.40 ± 1.31
KE	2.44 ± 0.53	16.66 ± 6.62	15.76 ± 4.84	10.17 ± 2.63	9.52 ± 2.20	7.55 ± 1.70
APF	1.04 ± 0.19	7.27 ± 3.07	6.48 ± 2.18	4.14 ± 1.00	4.04 ± 0.73	3.26 ± 0.61
ADF	0.49 ± 0.15	3.59 ± 1.53	3.25 ± 1.26	1.94 ± 0.68	1.89 ± 0.58	1.51 ± 0.46

Each time point RTD was calculated at all time points starting from onset at every 50 ms interval. The unit for maximum joint torques were “Nm/kg”, and RTDs were “Nm/kg/s”. HF, hip flexion; HE, hip extension; KF, knee flexion; KE, knee extension; APF, ankle plantarflexion; ADF, ankle dorsiflexion.

Table 3

Table 3. Coefficients of correlation based on a single variable linear correlation analysis between maximum joint torques and rate of torque development at each time point and maximum recoverable lean angle

		Maximum torque	$RTD_{0\text{-}50}$	$RTD_{0\text{-}100}$	$RTD_{0\text{-}150}$	$RTD_{0\text{-}200}$	$RTD_{0\text{-}250}$
HF	r	0.561 **	0.587 **	0.635 **	0.560 **	0.510 **	0.473 **
	ES	0.46	0.53	0.68	0.46	0.35	0.29
HE	r	0.301 *	0.142	0.407 **	0.285 *	0.329 *	0.345 **
	ES	0.10	0.02	0.20	0.09	0.12	0.14
KF	r	0.341 *	0.540 **	0.543 **	0.197	0.307 *	0.353 **
	ES	0.13	0.41	0.42	0.04	0.10	0.14
KE	r	0.237	0.237	0.128	0.095	0.141	0.132
	ES	0.06	0.06	0.02	0.01	0.02	0.02
APF	r	0.334 *	0.428 **	0.516 **	0.220	0.331 *	0.352 **
	ES	0.13	0.22	0.36	0.05	0.12	0.14
ADF	r	0.538 **	0.160	0.157	0.252	0.381 **	0.401 **
	ES	0.41	0.03	0.03	0.07	0.17	0.19

A significant correlation was denoted by *= $p < 0.05$, and **= $p < 0.01$.
HF, hip flexion; HE, hip extension; KF, knee flexion; KE, knee extension; APF, ankle plantarflexion; ADF, ankle dorsiflexion; ES, effect size given by Cohen's f^2

Table 4

Table 4. Result of multiple stepwise regression analysis for predicting maximum recoverable lean angle.

Variable	B	95% CI	SE	Beta	T	P
Model: R^2= 0.589, F= 27.27, p <0.001						
HF RTD_{0-50}	0.443	0.163-0.724	0.140	0.353	3.172	0.003
KF RTD_{0-100}	0.574	0.289-0.859	0.142	0.373	4.046	< 0.001
HF RTD_{0-150}	0.863	0.190-1.536	0.336	0.282	2.572	0.013

HF, hip flexion; KF, knee flexion; RTD, rate of torque development; B, unstandardized coefficients of B; 95%CI, 95% confidence interval for B and lower bound to upper bound; SE, standard error; Beta, standardized coefficients of Beta; T, t value; P, p value