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# Are Joint Torque Models Limited by an Assumption of Monoarticularity?

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This study determines whether maximal voluntary ankle plantar flexor torque could be more accurately represented using a torque generator that is a function of both knee and ankle kinematics. Isovelocity and isometric ankle plantar flexor torques were measured on a single participant for knee joint angles of 111° to 169° (approximately full extension) using a Contrex MJ dynamometer. Maximal voluntary torque was represented by a 19-parameter two-joint function of ankle and knee joint angles and angular velocities with the parameters determined by minimizing a weighted root mean square difference between measured torques and the two-joint function. The weighted root mean square difference between the two-joint function and the measured torques was 10 N-m or 3% of maximum torque. The two-joint function was a more accurate representation of maximal voluntary ankle plantar flexor torques than an existing single-joint function where differences of 19% of maximum torque were found. It is concluded that when the knee is flexed by more than 40°, a two-joint representation is necessary.

***Keywords:*** computer simulation, joint torque, biarticular muscle

Forward dynamics computer simulation models of human movement are typically either muscle driven or torque driven. In particular, muscle-driven models with various levels of complexity have been used to simulate a range of dynamic activities, such as long jumping (Hatze, 1981), vertical jumping (Pandy et al., 1990; van Soest et al., 1993) and drop jumping (Böhm et al., 2006) whereas torque-driven simulation models include long jumping (Alexander, 1990), vertical jumping (Cheng, 2008; Selbie and Caldwell, 1996), jumping for height (King et al., 2006), pole vaulting (Hubbard, 1980) and tumbling (King & Yeadon, 2004). Muscle-driven simulation models tend not to be specific to an individual whereas torque-driven models are typically customized to an individual using torque measurements (Yeadon & King, 2008).

Muscle-driven simulation models typically incorporate the force exerting characteristics of a muscle from force-velocity and force-length properties identified in the literature. These models can provide a valuable insight into the importance of specific characteristics of muscle structure and function in determining human movement. Typically these models are constructed using measurements sourced from a wide variety of experimental protocols, where the data may have come from animals, humans, in-vivo or in-vitro preparations, living participants or cadavers. Scovil and Ronsky (2006) have highlighted how perturbations to individual muscle model properties/parameters can introduce large errors to a simulation model.

Torque-driven simulation models can represent maximal voluntary torque at a joint as a nine-parameter torque–angle–angular velocity relationship that consists of a seven parameter torque–angular velocity relationship multiplied by a two-parameter quadratic torque–angle relationship (King et al., 2006). The torque–angular velocity relationship can be based on a tetanic Hill-type curve (Hill, 1938) multiplied by a differential activation function which expresses maximum voluntary activation as a function of angular velocity (Yeadon et al., 2006). The advantage of using a "torque generator" representation within a whole body simulation model is that the strength characteristics for each torque generator can be determined from subject-specific torque measurements using an isovelocity dynamometer (Yeadon et al., 2006; King et al., 2006).

While torque-driven simulation models can give an understanding of techniques used and the factors which affect optimal performance, one limitation is that the nine-parameter maximum voluntary joint torque representation does not accurately account for changes in the length and velocity of biarticular muscles during whole body movements. In particular, the maximum joint torque exerted by monoarticular muscles is a function of the angle and angular velocity of a single joint, while the maximum

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torque exerted by biarticular muscles is a function of two joint angles and angular velocities. As a consequence calculating maximum voluntary joint torque based upon the angle and angular velocity of a single joint may not be appropriate and could result in an over- or under-estimate of the maximum voluntary torque that could be produced during a simulation. For example, for ankle plantar flexion subject-specific joint torque characteristics are typically determined from maximum voluntary isovelocity ankle joint torque measurements with an extended knee (King et al., 2006). The resulting torque–angle–angular velocity relationship assumes that any changes in the knee joint angle do not affect maximum voluntary ankle plantar flexor torque. The soleus, gastrocnemius and plantaris are the three predominant ankle plantar flexors. The soleus is monoarticular crossing only the ankle joint while the gastrocnemius and plantaris are biarticular crossing both the knee and ankle joints. Using literature-derived values of physiological cross-sectional area (PCSA), pennation angle and moment arms the contributions of the two biarticular muscles can be up to 32% of the total plantar flexor torque (Table 1). As a consequence if the knee was fully flexed (biarticular muscles slack) the maximum voluntary ankle plantar flexor joint torque could be over-estimated by up to 32% if the changes in knee angle were not accounted for.

The purpose of this study was to incorporate changes in knee joint angle and angular velocity into a torque generator representation of ankle plantar flexor torque to compare the resulting subject-specific maximum voluntary joint torques with the torques calculated using a single-joint torque representation.

## Methods

Maximum voluntary ankle plantar flexor joint torques for a single male participant (height 1.78 m, mass 91 kg, age 36 yrs) with experience of using isovelocity dynamometers were measured at a variety of knee joint angles by varying the position of a rigid thigh support. The participant gave informed consent for the procedures which were carried out in accordance with a protocol approved by the Loughborough University Ethical Advisory Committee. Dynamometer torque, crank angle and crank velocity data were measured using a Contrex Multi-Joint isovelocity dynamometer (CMV AG, Switzerland). Planar knee joint angles were calculated from reflective markers placed over the lateral malleolus, the lateral collateral ligament of the knee and the greater trochanter of the femur which were tracked during each trial using a Vicon motion analysis system (OMG plc, UK) (Figure 1).

In particular the knee joint angle for each trial was calculated as the mean joint angle over the duration that the participant applied a plantar flexion torque, with full knee extension corresponding to an angle of 180°. The ankle joint angle (Figure 1) was calculated as the posterior angle between the midline of the shank and a line running from the ankle joint center to the 5th metatarsal and was determined directly from the crank angle data (with the sole of the foot perpendicular to the line of the shank the participant's ankle joint angle was 243°). Care was taken to ensure that joint and crank centers were aligned under load to limit any angular differences between joint and crank angles which can result from freedom in the system.

For each dynamometer trial the participant lay in a supine position, strapped firmly to the dynamometer at the feet, knees and hips to reduce unwanted movements. Following a warm-up the participant's maximum voluntary ankle plantar flexor torque was measured for seven ankle joint angles (isometric torque) and at a variety of concentric-eccentric isovelocities (50°/s, 100°/s, 150°/s, 200°/s, 250°/s and 300°/s) at each of five different knee joint angles (169°, 157°, 141°, 136° and 111°). The seven isometric torque measurements were made before measuring dynamic torques following which the knee angle was changed, starting with the most extended knee joint

**Table 1 Contributions to ankle plantar flexion torque**

| Muscle | Moment Arm (mm) | PCSA (mm$^2$) | Pennation Angle (°) | Contribution (%) |
|---|---|---|---|---|
| Gastroc. (medial)[a] | 52.8 | 4,177 | 14 | 21.1 |
| Gastroc. (lateral)[a] | 52.8 | 1,990 | 11 | 10.2 |
| Plantaris[a] | 52.8 | 209 | 4 | 1.1 |
| **Percentage of contribution from biarticular muscles acting about the knee and ankle: 32.4** | | | | |
| Soleus[a] | 52.8 | 11,868 | 26 | 55.4 |
| Flexor hallucis longus[b] | 26.6 | 1,408 | 17 | 3.5 |
| Flexor digitorum longus[b] | 23.0 | 991 | 11 | 2.2 |
| Tibialis posterior[a] | 8.0 | 3,622 | 17 | 2.7 |
| Peroneus longus[a] | 12.8 | 2,144 | 10 | 2.7 |
| Peroneus brevis[a] | 9.9 | 1,154 | 8 | 1.1 |
| **Percentage of contribution by monoarticular muscles (any plantar flexor not acting about the knee joint): 67.6** | | | | |

[a]Data from Klein et al. (1996)

[b]Data from Murray et al. (1976).


**Figure 1** — Data collection set-up showing participant positioning and joint angle definitions.

angle and ending with the most flexed. Each isometric and isovelocity trial were recorded once with a recovery period of 60 s between trials, 5 min while changing the knee angle and 1 hr after data had been collected on the first three knee angles. Each isovelocity trial consisted of two repetitions of concentric-eccentric movement with the participant asked to work maximally throughout the entire trial. The ranges of motion at the ankle and knee were maximized based upon the ranges of motion that the subject was comfortable with and the capabilities of the isovelocity dynamometer.

Torque and angle data were low pass filtered at 8 Hz using a fourth-order zero-lag Butterworth filter. Peak torque and corresponding ankle angle were identified for each isometric trial. Torque was sampled for periods of constant crank velocity, calculated from crank angle data, and then interpolated to provide a measure of torque at intervals of 1° throughout the isovelocity range (Yeadon et al., 2006). Corrections were made to the raw torque data, to account for passive torques resulting from the weight of the limbs and dynamometer attachments (Pavol & Grabiner, 2000). The resulting data consisted of maximal voluntary isometric and isovelocity ankle plantar flexion torque for the five knee angles, and these data were used to determine the 19 parameters for a two-joint torque generator function of ankle plantar flexion torque.

The 19-parameter two-joint torque generator function included both monoarticular and biarticular representations and expressed maximum voluntary ankle plantar flexion torque as a function of ankle angle $\theta_A$, knee angle $\theta_K$ and the corresponding two angular velocities $\omega_A$ and $\omega_K$. The 19-parameter function consisted of the sum of a nine parameter monoarticular function of $\theta_A$ and $\omega_A$ (Table 2; King et al., 2006), and a ten-parameter biarticular function of $\theta_A$, $\theta_K$, $\omega_A$ and $\omega_K$. The ten-parameter biarticular function was based on the nine-parameter function with one additional parameter R for the ratio of moment arms ($d_A$ and $d_K$) at the ankle and knee ($R = d_K / d_A$). This extra parameter allowed $\theta_A$ to be added to $\theta_K$ in a meaningful way so that the combined angle represented the "length" of the biarticular component $\theta_{Bi}$, where $\theta_{Bi} = \theta_A + R\theta_K$. In the same way the ratio R was used to allow $\omega_K$ to be added to $\omega_A$ to give a biarticular component angular velocity $\omega_{Bi}$, where $\omega_{Bi} = \omega_A + R\omega_K$. This resulted in a nine-parameter biarticular function of $\theta_{Bi}$ and $\omega_{Bi}$ (see appendix for a full description of the torque function). As a consequence the total ankle plantar flexion torque $T = T_9(\theta_A, \omega_A) + T_{10}(\theta_A, \theta_K, \omega_A, \omega_K)$ where $T_9(\theta_A, \omega_A)$

**Table 2 Calculated single and two joint torque parameters**

| | Two Joint | | | |
|---|---|---|---|---|
| **Parameter** | **Monoarticular** | **Biarticular** | **Single Joint** | **Bounds (LB–UB)** |
| $T_{max}$ | 270.06 | 172.68 | 376.68 | 1.4 ($T_0$) |
| $T_0$ | 192.90 | 123.34 | 269.05 | 0.4 to 1.6 ($T_0$) |
| $\omega_{max}$ | 16.76 | 18.31 | 20.26 | 7.5 to 30 |
| $\omega_c$ | 5.40 | 5.10 | 4.45 | 0.15 to 0.5 ($\omega_{max}$) |
| $k_2$ | 0.50 | 0.59 | 0.85 | 0.2 to 2.0 |
| $\theta_{opt}$ | 4.98 | 6.41* | 4.89 | 4.0 to 5.2 |
| $a_{min}$ | 0.71 | 0.84 | 0.7 | 0.2 to 1.0 |
| m | 0.06 | 0.04 | 0.06 | 0.0 to 0.8 |
| $\omega_1$ | –0.12 | –0.40 | –0.11 | –0.5 to 3.0 |
| R | — | 0.34 | — | 0.2 to 0.7 |

*Note.* Nomenclature and bounds based upon the following: maximum eccentric torque $T_{max}$ (N·m) (Dudley et al., 1990; Webber & Kriellers, 1997), maximum isometric torque $T_0$ (N·m), maximum concentric velocity $\omega_{max}$ (rad/s) (King et al., 2006), vertical asymptote ($\omega = -\omega_c$) $\omega_c$ (rad/s) (UB—Scovil and Ronsky, 2006); (LB—Umberger et al., 2006), width of torque–angle relationship $k_2$, optimum angle, $\theta_{opt}$ (rad) (UB permitted outside joint range where curve may be ascending only), minimum muscle activation, $a_{min}$, activation rate, m and point of inflection $\omega_1$ (rad/s) (Amiridis et al., 1996) and moment arm ratio R (LB—Brindle et al., 2008), (UB—Grieve et al., 1978). No boundary values were met for the optimal set of parameters for the single joint and two joint solutions.

*Biarticular bounds for $\theta_{opt}$ are 3.5–8.7.

represents the monoarticular torque and $T_{10}(\theta_A, \theta_K, \omega_A, \omega_K)$ represents the biarticular torque.

To determine the 19 subject-specific torque parameters an unbiased weighted root mean square (RMS) difference between the calculated joint torque and the measured isometric and isovelocity torque data were calculated as in equation (1) and was minimized using the simulated annealing algorithm (Corana et al., 1987). An unbiased RMS was calculated to account for the degrees of freedom (Carnahan et al., 1969). An initial estimate of the nine monoarticular parameters was obtained by minimizing the difference between the nine parameter function and the measured torque for a knee angle of 111° since with the knee in this flexed position the biarticular muscles cannot produce much ankle plantar torque (Muraoka et al., 2005). The initial parameter estimates for monoarticular parameters were allowed to vary by ±5% when optimizing to determine both monoarticular and biarticular parameters. All parameters were given upper and lower bounds and where possible these were based on physiologically realistic values found in the literature (Table 2).

Unbiased weighted RMS difference function =

$$\sqrt{\frac{n+m}{n+m-f}} \times \sqrt{\frac{w_1 \sum_{i=1}^{n} x_i^2 + w_2 \sum_{j=1}^{m} y_j^2}{nw_1 + mw_2}} \qquad (1)$$

For data points where the measured torque was greater than the function value, $w_1 = 100$, n = the number of data points, $x_i$ = difference between measured torque and the function value. And likewise for data points where the measured torque was less than the function value, $w_2 = 1$, m = the number of data points, $y_j$ = difference between measured torque and the function value, and f = number of function parameters (9 or 19). Using a weighted RMS difference resulted in a 19-parameter subject-specific function that represented maximum voluntary joint torque rather than the average torque produced by encouraging the function to give a better agreement with larger torque measurements. Weightings of 100 and 1 had been identified from fitting both single-joint and two-joint functions to pseudo torque data sets with added random noise representative of torque measurement errors. Typically the torques measured on the participant are more likely to be submaximal than supramaximal. The selected combination of weightings consistently provided torque representations with approximately 15% of torque measurements larger than calculated torques.

In addition the nine parameters for a single-joint torque generator function (King et al., 2006) were determined by minimizing the difference function in Equation (1) for the measured ankle plantar flexion torque data with the participant's knee in its most extended position (169°). A comparison of the parameters for the two functions was then made along with a comparison of maximal voluntary ankle plantar flexion torque for two specific sets of joint angles and velocities obtained from the literature (squat jumping; Kurokawa et al., 2001) and for drop landing (Yeadon et al., 2010).

## Results

The calculated 19 subject-specific two-joint parameters (Table 2) resulted in a torque generator function with a weighted RMS difference between the measured torques and those calculated by the function of 10 N·m (3% maximum torque) with 13% of the torque measurements greater than the calculated ones (Figures 2 through 5). For the torque measurements that were greater than the calculated values the RMS difference was 2%, while this difference was 8% for measured torques that were less than calculated torques. The two-joint representation closely matched the measured torques across the full range of knee joint angles with a range of 2–7% difference between calculated and measured torques for each of the different knee angles (Tables 3 and 4). Repeat measurements of ankle plantar flexor torque collected at the end of the session showed no signs of participant fatigue. At 50°/s the isovelocity range of ankle angles was approximately 45° and at 300°/s this had reduced to approximately 8°.

The calculated single-joint nine parameter function gave a weighted RMS difference between calculated torques and measured torques at a knee angle of 169° of 12 N·m (4% maximum torque) with 16% of torque measurements greater than the calculated values (Figure 6). Where the torque measurements were larger than the calculated torques the RMS difference was 2%, while this difference was 11% for measured torques that were less than calculated torques. The unbiased weighted RMS difference between the single-joint torque generator representation and experimental torque measurements collected at other knee angles showed much larger variations than the equivalent comparison with the two-joint torque function along with differences of 7–20% compared with 2–7% (Table 3).

Using the calculated torque parameters with an ankle velocity of –300°/s (the maximum used in this study), a knee angle of 169°, and a realistic maximum ankle joint angle of 263° (20° of dorsiflexion) resulted

**Table 3 Unbiased weighted RMS percentage differences between calculated torques and measured torques**

| Knee Angle | Two-Joint Torque Generator (%) | Single-Joint Torque Generator (%) |
|---|---|---|
| 169°[a] | 3.6 | 4.4 |
| 157° | 2.9 | 7.1 |
| 141° | 3.7 | 12.5 |
| 136° | 6.6 | 20.0 |
| 111° | 2.0 | 8.9 |

[a]Data set used to obtain single-joint model parameters.

**Table 4 Unbiased weighted RMS percentage differences between calculated torques (two-joint representation) and measured torques for different ankle angles and ankle angular velocities, and all knee angles**

| Angle | Difference (%) | Velocity | Difference (%) |
|---|---|---|---|
| 220° | 3.2 | –300°/s | 2.5 |
| 230° | 3.4 | –250°/s | 2.1 |
| 240° | 3.0 | –200°/s | 0.9 |
| 250° | 2.9 | –150°/s | 1.9 |
| | | –100°/s | 7.9 |
| | | –50°/s | 3.9 |
| | | 0°/s | 3.7 |
| | | 50°/s | 6.6 |
| | | 100°/s | 4.8 |
| | | 150°/s | 2.2 |
| | | 200°/s | 3.2 |
| | | 250°/s | 2.3 |
| | | 300°/s | 2.2 |

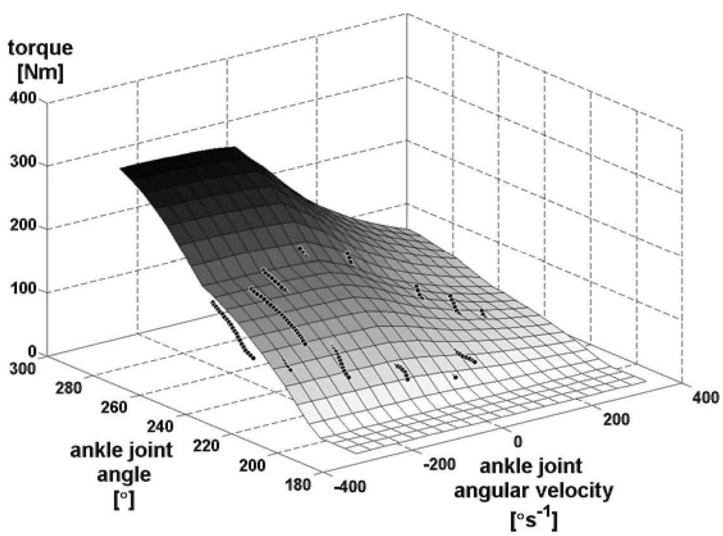


**Figure 2** — Two-joint torque generator fit to measured torques at the most flexed knee position of 111°

in a calculated maximal voluntary torque of 261 N·m (two-joint representation) compared with 243 N·m for the single joint representation (Table 5). Furthermore using the torque parameters and comparing the calculated ankle plantar flexion torques for realistic kinematic jump data (Table 5) demonstrated both agreement and disagreement between the two torque functions. The torques calculated using the kinematics from the squat jump at the instant of maximum ankle and knee joint angular velocities were small and less than 3 N·m different. Conversely calculating the ankle plantar flexion torques using the kinematics of a drop landing at the point of touch-down showed a large difference between the calculated torques from the single-joint representation and the two-joint representation (Table 5). The two-joint representation calculated 37 N·m less torque than the single-joint representation and this equates to 79% of the single-joint calculated torque.

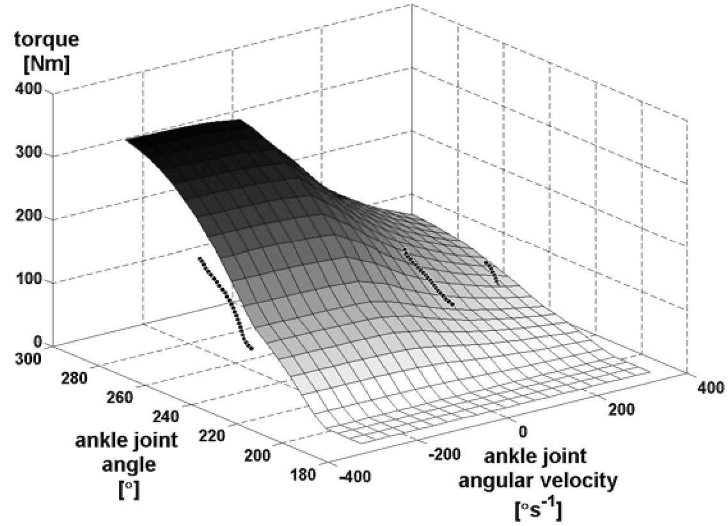


**Figure 3** — Two-joint torque generator fit to measured torques at the most extended knee position of 169°.

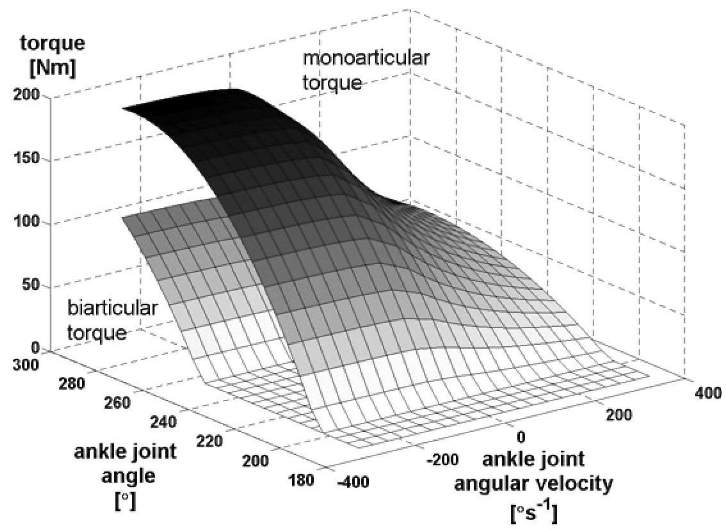


**Figure 4** — Monoarticular and biarticular torque contributions with a flexed knee.

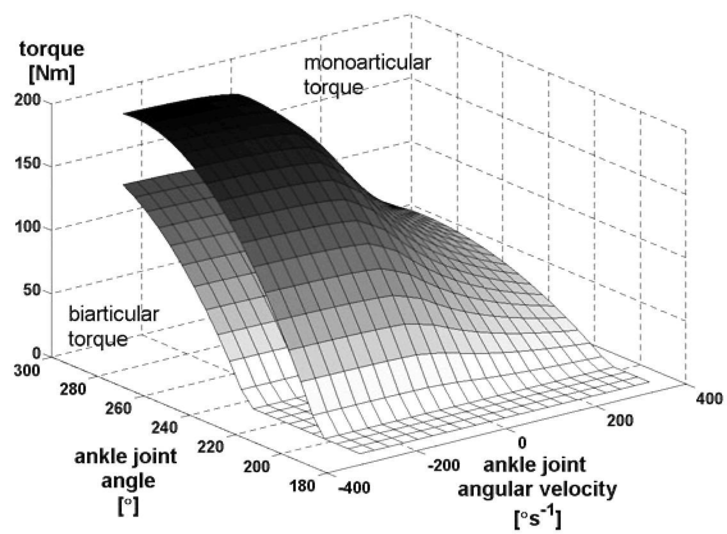


**Figure 5** — Monoarticular and biarticular torque contributions with an extended knee.

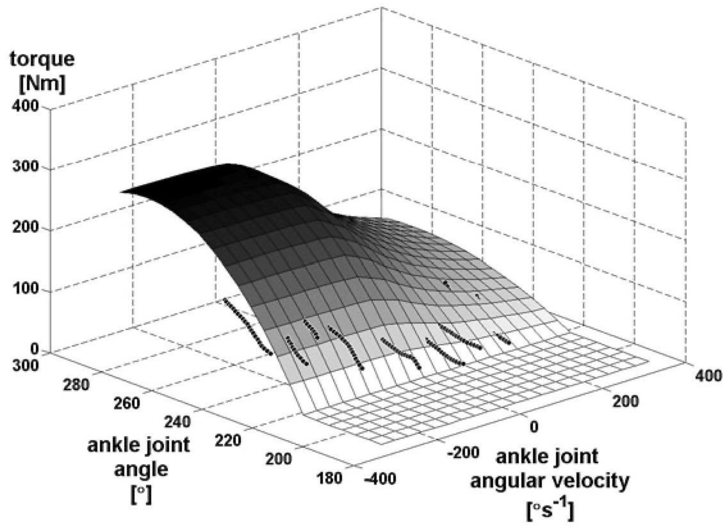


**Figure 6** — Single-joint torque generator surface and measured torques with a flexed knee.

**Table 5 Maximal voluntary ankle plantar flexor torques for three different sets of ankle / knee kinematics**

| | Ankle Angular Velocity (°/s) | Ankle Angle (°) | Knee Angular Velocity (°/s) | Knee Angle (°) | Single-Joint Calculated Torque (N·m) | Two-Joint Calculated Torque (N·m) |
|---|---|---|---|---|---|---|
| a | –300 | 263 | 0 | 169 | 243 | 261 |
| b | 733 | 229 | –710 | 149 | 10 | 7 |
| c | –7.3 | 245 | 117 | 133 | 173 | 136 |

[a]Maximum knee angle used, a realistic maximum ankle angle.

[b]Data from squat jump 25 ms before take-off and point of maximum ankle plantar flexion and knee extension velocities; Kurokawa et al. (2001).

[c]Data from drop landing, at point with heel contact and maximum posterior ankle joint angle and coinciding with highest ankle torques; Yeadon et al. (2010).

## Discussion

Although similar differences of 3–4% between calculated torques and measured torques used to determine function parameters existed for both the two-joint and single-joint representations, the single-joint model was found to provide little agreement with torques measured for the four knee joint angles not used to determine the parameters (Table 5). The unbiased difference between the single-joint torque representation and the measured torques with a flexed knee increased by up to a multiple of five (Table 3), while the two-joint representation gave a consistent and in all cases smaller difference across the full range. Therefore while accounting for any biases in the number of parameters the two-joint approach was able to provide a closer fit to the measured torques.

The single-joint representation in this study could not fit torque measurements collected across a wide range of knee angles whereas the two-joint representation could. Such results confirm that, as expected, changes in knee angle will affect biarticular muscle length and hence the exerted muscle force. When the knee is most flexed the biarticular torque in this study is smallest and this is in agreement with other literature findings which identify maximum biarticular muscle torques at greatest muscle lengths (Mohamed et al., 2002). The two-joint torque representation identifies that the participant's maximum voluntary isometric torque should occur with a fully dorsiflexed ankle and the knee fully extended. A number of studies have attempted to identify a relative contribution of the various lower limb muscles to ankle plantar flexor torque. In this study the maximum ankle plantar flexion torque contribution from the biarticular component totals 31% of total torque, which falls within the range of values found in the literature of 15–50% (Fugl-Meyer et al., 1980; Sale et al., 1982; Cresswell et al., 1995) and corresponds well to the estimate of 32% which can be derived from physiological measurements sourced from the literature (Table 1).

The ratio R of moment arms at the primary and secondary joints had an optimal value of 0.34 which compares well to values of 0.33 and 0.23 derived from passive ultrasound measurements (Brindle et al., 2008) and the ratio of 0.37 used for models of jumping by van

Soest and colleagues (van Soest et al., 1993; van Soest & Bobbert, 1993) derived from the original equations of Grieve et al. (1978). Measurements made of the moment arms of various lower limb muscles using tendon excursion methods and ultrasound (Grieve et al., 1978; Spoor et al., 1990; Klein et al., 1996) have found moment arm to change as a function of joint angle for muscles of the lower leg. The nature of how the moment arm changes with joint angle differs between studies: increasing with an increasing joint angle to the reverse. The current approach avoids assigning any relationship governing moment arm length with joint angle, opting instead for a single ratio. While such a method could introduce errors when moment arms change by a large amount, evidence from the study of Out et al. (1996) has shown that a simple mean may offer a suitable representation. In their sensitivity analysis of a muscle model for plantar-flexion they used morphological data in the literature to identify the mean moment arms of the triceps surae at ankle and knee joints. The two data sets used reported very different moment arm / joint angle relationships. In one the moment arm increased with joint angle and in the other it decreased. The authors demonstrated that the use of a single mean value for the moment arms for both data sets resulted in torque-angle relationships that were almost identical and this lends support to the method used in the current study.

The advantages of using a two-joint torque representation can be seen from calculations of maximal plantar flexion torque using literature sourced kinematic data of a squat jump and a drop landing. While the kinematic data are not subject-specific, the general kinematics are likely to be similar for different individuals and enabled the authors to speculate whether a two-joint representation might be appropriate for different applications. The two-joint representation, unlike the single-joint approach, was able to show that for the predominantly extended knee and ankle joint angles of the squat jump the single-joint and two-joint representations provide good agreement of the torques being exerted at the ankle joint. For the drop landing however, the biarticular muscles acting about the knee and ankle were shorter than their functional torque exerting lengths and as a consequence the single-joint representation calculated a much larger torque than the two-joint approach since the two-joint representation shows that only the monoarticular component was capable of exerting a torque.

Authors of previous studies which used single-joint torque generators to model dynamic tasks such as tumbling or running jumps (Yeadon & King, 2002; King et al., 2006) have proposed that not accounting for the effect of two-joint kinematics on biarticular muscle forces might be a potential source of error in their simulation models. In practice, however, for both of these studies the single-joint models gave good agreement with the participants' performance. Two possible explanations exist. The first is that the joint angles and velocities where maximum torque is used correspond closely to the dynamometer torque measurement joint orientations: for example, the knee is almost fully extended when the model is called upon to exert a maximal ankle plantar flexor torque. The current study found that for a knee flexed by less than approximately 40° from full extension and exerting a maximal plantar flexor torque, the difference between a two-joint representation and a single-joint representation would be less than 12%. The second explanation is that for these models the body had a high initial whole body momentum and the joint torque generators were called upon to make relatively small contributions to modify the whole body momentum. As a consequence the difference between the model activation time histories of the torque generators and those of the participant may be very small. In contrast the performance of a squat jump involves much greater flexion of the knee and a stationary starting posture means that the whole body angular momentum is generated entirely by the torques at each joint. For activities such as these the use of a two-joint representation of torque is likely to be of greater importance than for the previous simulation models of tumbling and running jumps (King & Yeadon, 2004; King et al., 2006).

The measurements of ankle plantar flexor torque for a single participant provided the data for this study. Most experimental studies require multiple participants to establish statistical significance and hence to speculate upon the relationship between a hypothesized cause and effect. For this study the mechanics are well understood; the existence of biarticular muscles and a muscle force-length relationship are not being questioned and so a single subject is appropriate. This study has shown that the simplification of a two-joint relationship into a single-joint relationship can in some cases limit the accuracy of a subject-specific torque generator model.

The findings from this study have implications for the design of whole body torque-driven simulation models. By taking into account the length changes of biarticular muscles it is possible to overcome a potential limitation of single-joint torque generator models while building on the strength of determining subject-specific parameters from measurements on the subject without requiring data from the literature. To generalize the findings of this study to other subjects would require a lot of subject-specific data in order that scaling from a few subject-specific measurements would be possible. Clearly there is more work to be done to adapt this methodology for use at other joints in the body which have a different make-up of monoarticular and biarticular muscles. In conclusion, it is clear from this study that it is possible to determine subject-specific parameters for a two-joint torque generator representation from measurements on the subject. The two-joint representation offers more accuracy than a single-joint torque generator approach for calculating maximum voluntary ankle plantar flexor torques as a function of ankle and knee angles, and is especially important when the knee is flexed by more than approximately 40°.

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# Appendix

For the following description "component" refers to the monoarticular or biarticular element properties. The torque for any component velocity ω and component angle θ combination was derived from the tetanic torque $T_{(4)}$ represented by four free parameters, multiplied by activation a, the fraction of maximum activation at each velocity, and then multiplied by a fraction of the maximum torque available at any velocity for a given component angle and governed by a quadratic torque-angle function. Hence: $T = T_{(4)}\, a\, T_\theta$.

The component angle θ and velocity ω for the monoarticular component are equal to the ankle joint angle $\theta_A$ and ankle joint velocity $\omega_A$ respectively, while for the biarticular component a ratio of moment arms R enables the ankle and knee joint angles to be summed such that $\theta = \theta_A + R\theta_K$. The biarticular component velocity is derived similarly from the addition of ankle and knee velocities $\omega_A$ and $\omega_K$, using $\omega = \omega_A + R\omega_K$. The monoarticular and biarticular torques were represented by the following functions (Yeadon et al., 2006):

For concentric velocities:

$$T_{(4)} = \frac{C}{(\omega_c + \omega)} - T_c$$

$$\text{where } T_c = \frac{T_0 \omega_c}{\omega_{max}} \quad \text{and} \quad C = T_c(\omega_{max} + \omega_c)$$

For eccentric velocities:

$$T_{(4)} = \frac{E}{(\omega_e - \omega)} - T_{max}$$

$$\text{where } \omega_e = \frac{(T_{max} - T_0)\omega_{max}\omega_c}{kT_0(\omega_{max} + \omega_c)},$$

$$E = -(T_{max} - T_0)\omega_e \quad \text{and} \quad T_{max} = 1.4T_0$$

Activation a for both concentric and eccentric velocities was derived from a sinusoidal function of the velocity-activation relationship:

$$\omega - \omega_1 = \frac{m(a - \tfrac{1}{2}(a_{min} + a_{max}))}{(a_{max} - a)(a - a_{min})}$$

The torque at any angle was calculated using a quadratic torque-angle function (King et al., 2006): $T_\theta = 1 - k_2(\theta - \theta_{opt})^2$.