

## Original Article

### **An important role of the biarticular hamstrings is to exert internal/external rotation moments on the tibia during vertical jumping**

Daniel J Cleather

School of Sport, Health and Applied Sciences, St. Mary's University, Twickenham, UK

Corresponding author:

Daniel Cleather,

St. Mary's University,

Waldegrave Road,

Twickenham.

TW1 4SX

UK

Tel: +44 7973 873 516

Email: [dancleather@hotmail.com](mailto:dancleather@hotmail.com) / [daniel.cleather@stmarys.ac.uk](mailto:daniel.cleather@stmarys.ac.uk)

#### **Published as:**

Cleather, D. J. (2018). An important role of the biarticular hamstrings is to exert internal/external rotation moments on the tibia during vertical jumping. *Journal of Theoretical Biology*, 455, 101-108. doi: [10.1016/j.jtbi.2018.07.013](https://doi.org/10.1016/j.jtbi.2018.07.013)



## **Abstract**

Most research considering biarticular muscle function has tended to focus on the sagittal plane. Instead, the purpose of this study was to evaluate the internal/external rotation moment arms of the biarticular muscles of the knee, and then to explore their function. The FreeBody musculoskeletal model of the lower limb was used to calculate the moment arms and moments that each of the muscles of the knee exerted on the proximal tibia of 12 athletic males during vertical jumping. Biceps femoris and tensor fascia latae were external rotators of the tibia, whereas semimembranosus, semitendinosus, sartorius, gracilis, popliteus and the patellar tendon were internal rotators. The magnitudes of the internal/external rotation and flexion moments exerted on the tibia by the biarticular hamstrings were similar, suggesting that the creation of internal/external rotation is a key aspect of their role. One potential reason is to stabilise the tibia during femoral extension (and it is argued that it may be helpful to characterise the creation of active joint stability as the stabilisation of one segment during the rotation of an adjacent segment). A second explanation may be to mechanically couple hip abduction when the hip is flexed with internal rotation of the tibia.

**Keywords:** FreeBody; musculoskeletal modelling; knee; biarticular muscles; moment arm



## INTRODUCTION

The ankle, knee and hip joints of the human lower limb are articulated by a mixture of monoarticular and biarticular muscles. The presence of the biarticular muscles in particular has attracted considerable attention from scientists who have sought to understand their role in movement. Theoretically the limb could be articulated simply by monoarticular muscles and this seems to suggest that there is a particular reason for the existence of the biarticular muscles. A number of theories as to this role have been advanced within the literature. Probably the most popular explanation is that the biarticular muscles permit work that is done by a muscle at one joint to then be expressed at a different joint (Van Ingen Schenau and Bobbert, 1993) and that this then brings with it a number of advantages. Similarly, it has been observed that the biarticular muscles do not require as great a length change during many movements meaning that the velocity of shortening is slower and thus allowing greater force production (Bobbert et al., 1986a, 1986b; Cleather et al., 2015). We have recently suggested that a key advantage of the biarticular hamstrings and gastrocnemius is that they both permit a greater involvement of the quadriceps in closed kinetic chain extension than would be possible if the lower limb had only monoarticular muscles (Cleather et al., 2015). Finally it has been suggested that the presence of the biarticular muscles promotes joint stability, by encouraging co-contraction of agonist and antagonist muscles at a joint (Baratta et al., 1988).

One thing that all of these explanations tend to have in common is that they are based predominantly on a consideration of the mechanics of movement in the sagittal plane. One contributing factor to this is the fact that the function of the muscles of the lower limb in the other anatomical planes has generally not received the same degree of attention within the literature. For instance, only one previous study has sought to quantify the internal/external



rotation moment arms of the muscles of the knee (Buford Jr et al., 2001). This in turn hampers efforts to understand the role of the biarticular muscles in 3 dimensions. The purpose of this study is therefore to use musculoskeletal modelling technology to address this limitation. Firstly, a model of the musculoskeletal geometry of the lower limb will be used to quantify the internal/external rotation arms of all of the muscles of the knee during vertical jumping. Secondly, the same model will be used to estimate the forces in the biarticular muscles in all 3 planes of motion during vertical jumping.

## **METHODS**

### **Experimental approach**

This study employs a musculoskeletal modelling approach in order to quantify the transverse plane moments applied to the tibia by the muscles of the knee during vertical jumping. In particular, a publicly available model of the lower limb (FreeBody; Cleather and Bull, 2015) is used to calculate the muscle and joint contact forces expressed by athletic men during vertical jumping using an optimization based inverse dynamics methodology (Cleather et al., 2011a, 2011b; Cleather and Bull, 2011a) and the moments are then derived from this analysis. Vertical jumping has frequently been employed as a model to study biarticular muscle function as it is a closed kinetic chain extension movement with a proximal to distal transfer of energy (Bobbert and van Ingen Schenau, 1988).

### **Subject characteristics**

This study is based upon data collected from 12 athletic men performing vertical jumping (see Table 1 for subject characteristics). The data used in this study has been presented previously in a number of our previous publications (Cleather et al., 2011a; Cleather and



Bull, 2010a, 2010b, 2011a, 2011b, 2015) and is publicly available as electronic supplementary material to our article “The development of a segment-based musculoskeletal model of the lower limb: Introducing FreeBody” (Cleather and Bull, 2015). The collection of this data set was approved the institutional review board of St Mary’s University College and all subjects provided informed consent.

Table 1. Subject characteristics.

	Mean $\pm$ SD
Number of Subjects	12
Age (years)	27.1 $\pm$ 4.3
Height (m)	1.786 $\pm$ 0.074
Body Mass (kg)	83.7 $\pm$ 9.9
Maximum Jump Height (m)	0.38 $\pm$ 0.05

## Instrumentation

Kinematic data consisting of the positions of 18 retro-reflective markers was collected using an 8 camera Vicon optical motion capture system (Vicon MX system, Vicon Motion Systems Ltd, Oxford, UK). The markers were placed on key anatomical landmarks of the right lower limb in accordance with our previous work (Table 2; Cleather and Bull, 2015). The ground reaction force of the right foot was recorded using a force plate (Kistler Type 9286AA, Kistler Instrumente AG, Winterthur, Switzerland) that was synchronised with the Vicon system. All data was collected at 200 Hz.



Table 2. Marker positions used for data capture.

Location
Calcaneus
Tuberosity of the fifth metatarsal
Head of the second metatarsal
1 additional marker placed on the foot
Apex of the lateral malleolus
Apex of the medial malleolus
3 additional markers placed on the shank segment
Lateral femoral epicondyle
Medial femoral epicondyle
3 additional markers placed on the thigh segment
Right anterior superior iliac spine
Left anterior superior iliac spine
Right posterior superior iliac spine
Left posterior superior iliac spine

## Procedure

Subjects first performed a standardised warm up, consisting of running drills, squats, lunges, and vertical jumps. Subjects then performed a series of maximal vertical jumps with their hands on their head. A minimum of 4 maximal jumps was recorded for each subject and then the highest vertical jump of each subject was then selected for analysis.

## Data analysis

This study employs the FreeBody musculoskeletal model of the lower limb that is freely available at [www.msksoftware.org.uk](http://www.msksoftware.org.uk). The development of FreeBody over the last 8 years has been described in extensive detail elsewhere (Cleather et al., 2011a, 2011b; Cleather and Bull, 2010a, 2011a, 2015). FreeBody has been extensively tested, including studies of its reliability (Price et al., 2017), validity (by comparison of its predictions to experimentally measured values; Cleather and Bull, 2015; Ding et al., 2016; Price et al., 2016) and its



sensitivity to some of the key assumptions on which it is based (Cleather and Bull, 2010a, 2010b, 2011b; Southgate et al., 2012) and it has been used to study the effect of exercise interventions on subsequent athletic performance (Czasche et al., 2017; Parr et al., 2017). To this end, only a brief description of the model is provided here, and the interested reader is directed to the previously cited publications.

The FreeBody lower limb model consists of 5 rigid segments representing the foot, calf, thigh, patella and pelvis. The position of the segments is determined from the motion capture data and then the musculoskeletal geometry is added based upon the data set of Klein Horsman et al. (Klein Horsman et al., 2007). The equations of motion in the global frame (Dumas et al., 2004) are established for each frame based upon this data, the anthropometry of the segments (de Leva, 1996) and the ground reaction force that is measured from the force plate. Solving the 22 equations of motion (Equation 1) provides the values of the 193 unknown muscle, ligament and joint contact forces. Clearly, the system is overdetermined as it has many more unknowns than equations, and so an optimization approach is taken to pick the solution that is considered to be most physiologically likely (Equation 2). The optimization process is performed using the `fmincon` function of MATLAB® (version 2016b; Mathworks, 1 Apple Hill Drive, Natick, MA, USA).



$$\begin{pmatrix}
\hat{p}_1^1 & \dots & \hat{p}_M^1 & \hat{p}_{pt}^1 & \hat{q}_1^1 & \dots & \hat{q}_N^1 & -I_{3 \times 3} & E_{3 \times 3} & E_{3 \times 3} & E_{3 \times 3} & E_{3 \times 3} \\
\hat{p}_1^2 & \dots & \hat{p}_M^2 & \hat{p}_{pt}^2 & \hat{q}_1^2 & \dots & \hat{q}_N^2 & I_{3 \times 3} & -I_{3 \times 3} & -I_{3 \times 3} & E_{3 \times 3} & E_{3 \times 3} \\
\hat{p}_1^3 & \dots & \hat{p}_M^3 & \hat{p}_{pt}^3 & \hat{q}_1^3 & \dots & \hat{q}_N^3 & E_{3 \times 3} & I_{3 \times 3} & I_{3 \times 3} & -I_{3 \times 3} & I_{3 \times 3} \\
\hat{r}_1^1 \times \hat{p}_1^1 & \dots & \hat{r}_M^1 \times \hat{p}_M^1 & \hat{r}_{pt}^1 \times \hat{p}_{pt}^1 & \hat{s}_1^1 \times \hat{q}_1^1 & \dots & \hat{s}_N^1 \times \hat{q}_N^1 & E_{3 \times 3} & E_{3 \times 3} & E_{3 \times 3} & E_{3 \times 3} & E_{3 \times 3} \\
\hat{r}_1^2 \times \hat{p}_1^2 & \dots & \hat{r}_M^2 \times \hat{p}_M^2 & \hat{r}_{pt}^2 \times \hat{p}_{pt}^2 & \hat{s}_1^2 \times \hat{q}_1^2 & \dots & \hat{s}_N^2 \times \hat{q}_N^2 & \tilde{d}^2 & -\tilde{h}_1^2 & -\tilde{h}_2^2 & E_{3 \times 3} & E_{3 \times 3} \\
\hat{r}_1^3 \times \hat{p}_1^3 & \dots & \hat{r}_M^3 \times \hat{p}_M^3 & \hat{r}_{pt}^3 \times \hat{p}_{pt}^3 & \hat{s}_1^3 \times \hat{q}_1^3 & \dots & \hat{s}_N^3 \times \hat{q}_N^3 & E_{3 \times 3} & \tilde{d}_1^3 & \tilde{d}_2^3 & E_{3 \times 3} & \tilde{f}^3 \\
\hat{p}_1^{pat} & \dots & \hat{p}_M^{pat} & \hat{p}_{pt}^{pat} & & & E_{3 \times N} & E_{3 \times 3} & E_{3 \times 3} & E_{3 \times 3} & E_{3 \times 3} & -I_{3 \times 3} \\
\rho_1 & \dots & \rho_M & -1 & & & E_{1 \times N} & E_{1 \times 3} & E_{1 \times 3} & E_{1 \times 3} & E_{1 \times 3} & E_{1 \times 3}
\end{pmatrix}
\begin{pmatrix}
F_1 \\
\vdots \\
F_M \\
F_{pt} \\
L_1 \\
\vdots \\
L_N \\
\hat{R}^1 \\
\hat{R}_1^2 \\
\hat{R}_2^2 \\
\hat{R}^3 \\
\hat{R}^{pat}
\end{pmatrix}$$

$$= \begin{pmatrix}
m^1(\hat{a}^1 - \hat{g}) - \hat{S}^0 \\
m^2(\hat{a}^2 - \hat{g}) \\
m^3(\hat{a}^3 - \hat{g}) \\
m^1 \hat{c}^1 \times (\hat{a}^1 - \hat{g}) + Y_{3 \times 3}^1 \ddot{\hat{\phi}}^1 + \dot{\hat{\phi}}^1 \times Y_{3 \times 3}^1 \dot{\hat{\phi}}^1 - \hat{d}^1 \times \hat{S}^0 - \hat{W}^0 \\
m^2 \hat{c}^2 \times (\hat{a}^2 - \hat{g}) + Y_{3 \times 3}^2 \ddot{\hat{\phi}}^2 + \dot{\hat{\phi}}^2 \times Y_{3 \times 3}^2 \dot{\hat{\phi}}^2 \\
m^3 \hat{c}^3 \times (\hat{a}^3 - \hat{g}) + Y_{3 \times 3}^3 \ddot{\hat{\phi}}^3 + \dot{\hat{\phi}}^3 \times Y_{3 \times 3}^3 \dot{\hat{\phi}}^3 \\
E_{3 \times 1} \\
0
\end{pmatrix}$$

... Equation 1

$$\min_{F_i, L_j} J = \sum_{i=1}^M \left( \frac{F_i}{F_{max_i}} \right)^3 + \sum_{j=1}^N \left( \frac{L_j}{L_{max_j}} \right)^3$$

... Equation 2

where:

- $\hat{a}^k$  linear acceleration of the centre of mass of segment  $k$
- $\hat{c}^k$  vector from centre of rotation of joint at proximal end of segment  $k$  to centre of mass of segment  $k$
- $\hat{d}^k$  vector from centre of rotation of the joint at the proximal end of segment  $k$

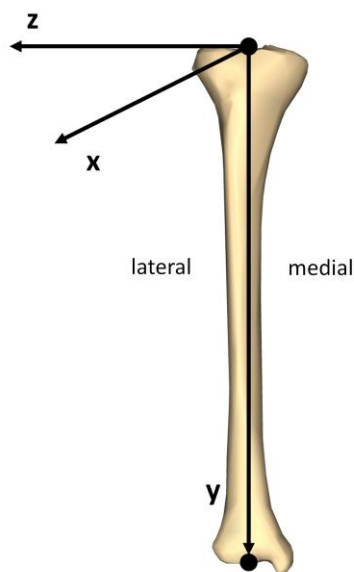


	to centre of rotation of the joint at the distal end of segment $k$
$\tilde{d}^k$	skew-symmetric matrix of vector $\tilde{d}^k$
$\tilde{d}_l^3$	skew-symmetric matrix of vector from centre of rotation of hip to tibiofemoral joint contact $l$ (there are two contact points thus $l=1$ or $2$ )
$E_{3 \times 3}$	$3 \times 3$ matrix of zeros
$\tilde{f}^3$	skew-symmetric matrix of vector from centre of rotation of hip to contact point of patella with the femur
$F_i$	magnitude of force in muscle $i$
$Fmax_i$	maximum possible force in muscle $i$ (upper bound)
$\hat{g}$	acceleration due to gravity
$\tilde{h}_l^2$	skew-symmetric matrix of vector from centre of rotation of knee to tibiofemoral joint contact $l$
$i$	muscle number
$I_{3 \times 3}$	$3 \times 3$ identity matrix
$j$	ligament number
$J$	cost function
$k$	segment number
$L_j$	magnitude of force in ligament $j$
$Lmax_j$	maximum possible force in ligament $j$ (upper bound)
$m^k$	mass of segment $k$
$M$	total number of muscles
$N$	total number of ligaments
$\hat{p}_i^k$	unit vector representing the line of action of force created by muscle $i$ that acts on segment $k$ (zero if muscle does not insert on segment $k$ )
$pat$	Patella
$pt$	patellar tendon
$\hat{q}_j^k$	unit vector representing the line of action of force created by ligament $j$ that acts on segment $k$ (zero if ligament does not insert on segment $k$ )
$\hat{r}_i^k$	vector from centre of rotation of joint at proximal end of segment $k$ to point of action of muscle $i$ on segment $k$ (zero if muscle does not insert on segment $k$ )
$\hat{R}^k$	vector representing $x$ , $y$ and $z$ components of reaction force acting at proximal end of segment $k$
$\hat{R}_l^k$	vector representing $x$ , $y$ and $z$ components of reaction force $l$ acting at proximal end of segment $k$
$\hat{s}_j^k$	vector from centre of rotation of joint at proximal end of segment $k$ to point of action of ligament $j$ on segment $k$ (zero if ligament does not insert on segment $k$ )
$-\hat{S}^k$	inter-segmental force acting on proximal end of segment $k$
$-\hat{W}^k$	inter-segmental moment acting on proximal end of segment $k$
$Y_{3 \times 3}^k$	inertia tensor of segment $k$
$\rho_i$	ratio of patella to quadriceps tendon forces for muscle $i$ (zero if the muscle is not part of the quadriceps muscle group)
$\dot{\phi}^k$	angular velocity of segment $k$
$\ddot{\phi}^k$	angular acceleration of segment $k$



An aspect of the FreeBody model that is particularly important to understand when interpreting the results of this study, is that the equations of motion are posed on a “segmental” basis as opposed to the more common “joint based” approach (Cleather et al., 2014; Cleather and Bull, 2012). This means that rotation effect of all of the individual force vectors acting upon a segment are modelled. The moments attributed to each muscle that are presented in this study are the moments of the muscle on the tibia about the origin of the tibia (Figure 1). All moments that are presented are given in the local coordinate system of the tibia.

Figure 1. Anterior view of the right tibia illustrating the local coordinate system of the tibia segment in FreeBody (image of the tibia was taken from [https://commons.wikimedia.org/wiki/File:Right\\_tibia\\_-\\_close\\_up\\_-\\_anterior\\_view.png](https://commons.wikimedia.org/wiki/File:Right_tibia_-_close_up_-_anterior_view.png) on 7<sup>th</sup> December 2017. The author and licensor of the image is “BodyParts3D, © The Database Center for Life Science licensed under CC Attribution-Share Alike 2.1 Japan”).



## Data Analysis



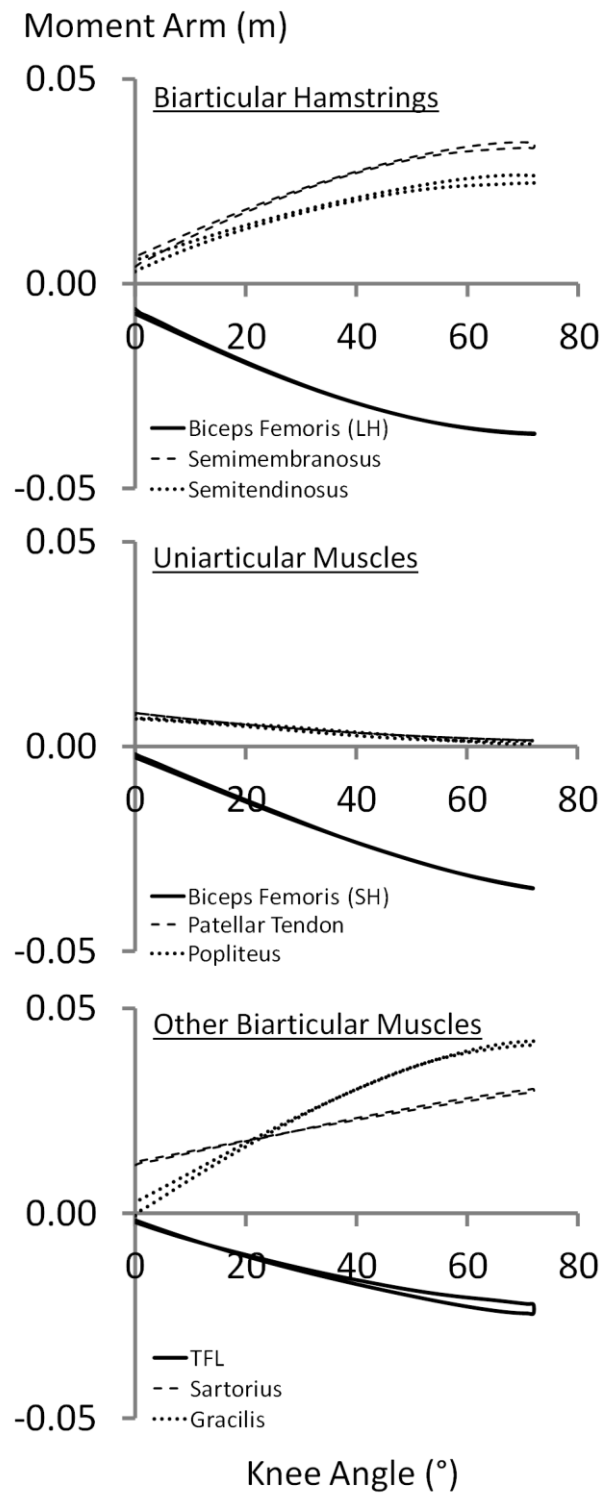
Firstly, the moment arms of biceps femoris (long and short heads), semitendinosus, semimembranosus, tensor fascia latae (TFL), gracilis, sartorius, popliteus and the patellar tendon were calculated for each jump. The moment arm by knee flexion angle curve was then spline interpolated using MATLAB to find the moment arm at each degree of knee flexion from 0° to 72° of knee flexion (72° being the smallest peak knee flexion angle reached by the participants during vertical jumping). The mean moment arm at each knee flexion angle was then calculated to produce composite mean curves of the moment arm/knee flexion angle relationship during vertical jumping. Next the mean moments impressed by each of the above muscles on the tibia were normalised relative to the peak knee flexion angle. These curves were then spline interpolated in order to produce composite mean curves of the muscle moments relative to the normalised knee flexion angle.

## **RESULTS**

The internal/external rotation moment arm of all muscles increased with increasing knee flexion except for popliteus and the patellar tendon (Figure 2). Biceps femoris (long and short heads) and TFL were external rotators of the proximal tibia. Semimembranosus and semitendinosus, sartorius, gracilis, popliteus and the patellar tendon were all internal rotators of the proximal tibia.



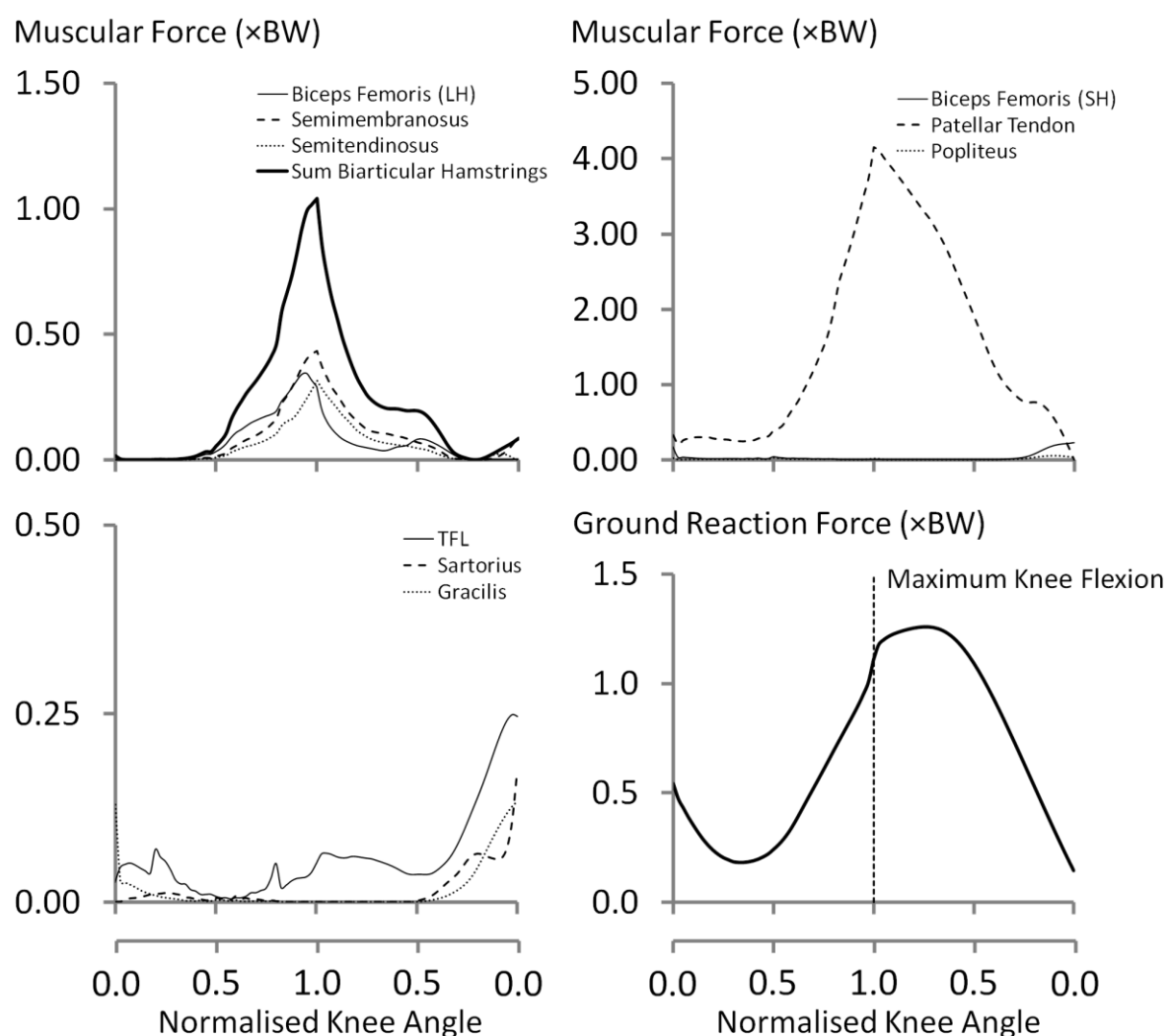
Figure 2. Transverse plane moment arms of muscles acting on the proximal tibia as a function of knee flexion angle during vertical jumping (mean internal (+)/external (-) rotation moment arm of the tibia for 12 subjects).





The ground reaction force peaked just after the point of maximum knee flexion (Figure 3). In contrast, the biarticular hamstrings and quadriceps (i.e. the force in the patellar tendon) were most active around the point of maximum knee flexion. TFL, sartorius, gracilis, popliteus and the short head of biceps femoris were more active near take-off.

Figure 3. Mean predictions of muscular forces and measured ground reaction forces for 12 subjects during vertical jumping.



The patellar tendon was the strongest rotator of the tibia, creating an extension moment that was 10 times greater than any other moment presented here (Figure 4). The patellar tendon also impressed modest internal rotation and abduction moments on the tibia. Aside from the



action of the patellar tendon, the biarticular hamstrings were the strongest rotators of the tibia. In contrast to the patellar tendon, the magnitude of the individual internal/external rotation moments created by each of the long head of biceps femoris, semitendinosus and semimembranosus were similar to the flexion moments they created. The other muscles acting on the proximal tibia had smaller effects that were predominantly seen just before take-off. Finally, the period when there was an inter-segmental internal rotation moment acting on the proximal tibia coincided with the period when the net effect of the biarticular hamstrings was to create an internal rotation of the proximal tibia (Figure 5).



Figure 4. Mean moments applied to the tibia during vertical jumping (mean abduction (+)/adduction (-), internal (+)/external (-) rotation and extension (+)/ flexion (-)).

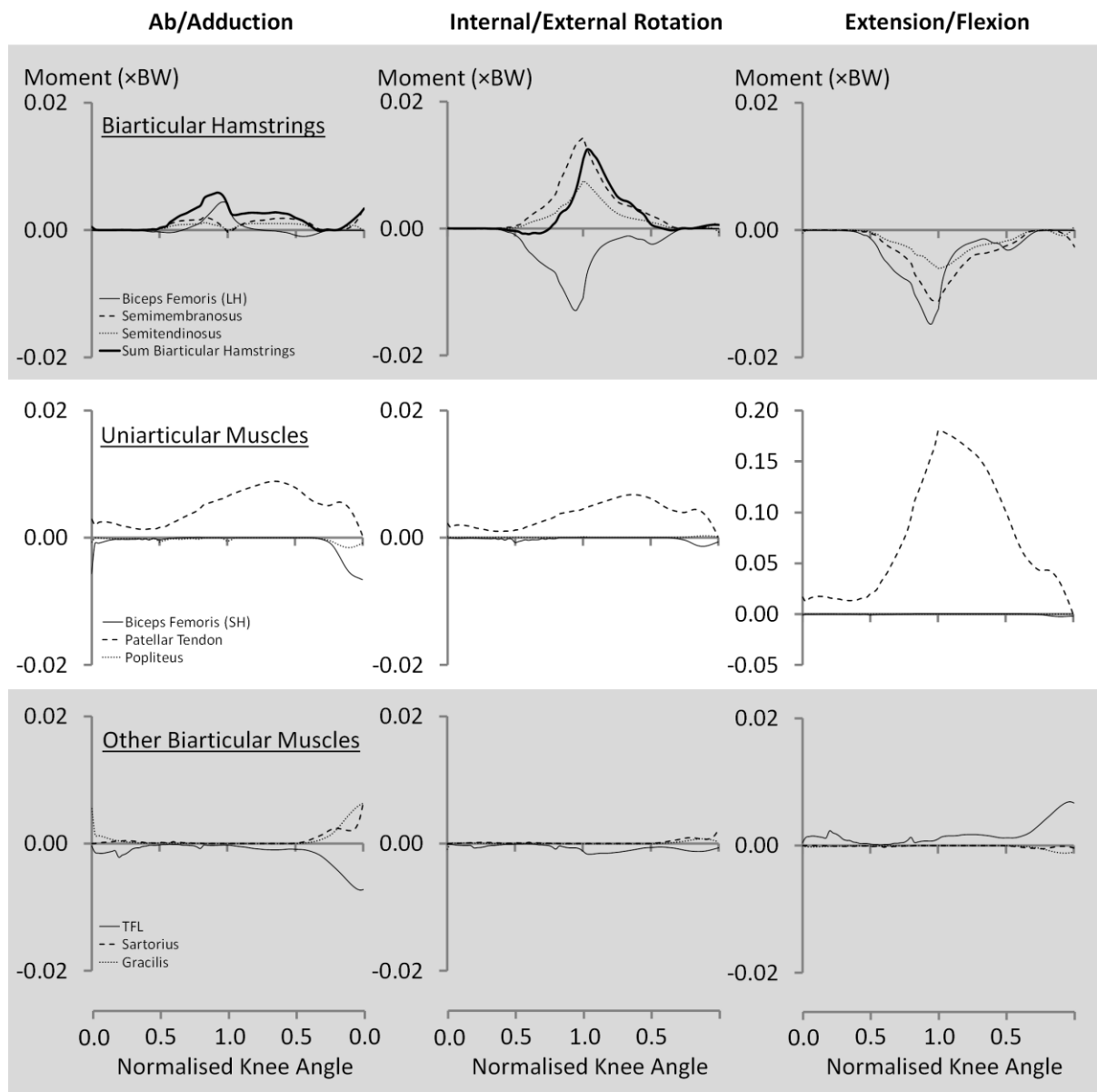
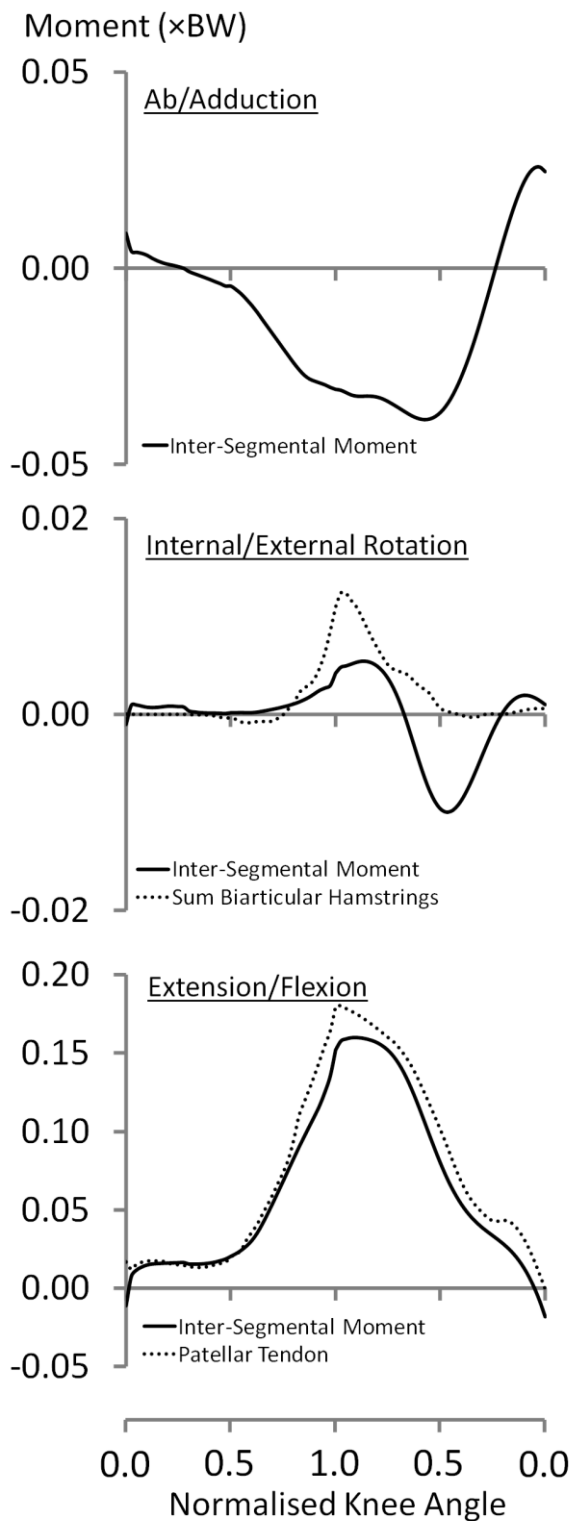




Figure 5. A comparison of the mean inter-segmental knee moments to selected muscular moments acting upon the tibia (mean abduction (+)/adduction (-), internal (+)/external (-) rotation and extension (+)/ flexion (-)).





## DISCUSSION

### Comparison of Model Outputs to Previous Literature

It should be noted that the moment arms presented in this work are based upon the cadaveric data set collected by Klein Horsman and colleagues (2007) which has then been geometrically scaled to the anthropometry of each subject. This in turn means that the moment arms presented here are estimates – they have not been directly measured for each subject. Despite this however, the moment arms presented in Figure 2 are generally consistent with previous results reported in the literature. For instance, imaging of the patellar tendon during weight bearing knee flexion has shown that the tendon is orientated medially in the coronal plane when the knee is extended (that is, it will create an internal rotation of the tibia), and that this angle diminishes with increasing knee flexion, such that at 90° of knee flexion it is orientated almost vertically (DeFrate et al., 2004). These geometrical observations are in accord with the decreasing internal rotation moment arm of the patellar tendon found in this study. There is a paucity of previous work that has sought to specifically quantify the internal/external rotation moment arms of the muscles of the knee. One notable exception is the cadaver study of Buford and colleagues (Buford Jr et al., 2001). The trends reported in the present study are remarkably similar to Buford's work – that is, the direction of internal or external rotation is the same for all muscles and the direction of the change with increasing knee flexion is also largely in agreement.

Similarly, it is also important to understand that the muscle forces that are reported here are also estimates. In particular, the equations of motion describing the movement have many solutions, and the most “optimal” solution is chosen by minimising the objective function presented in Equation 2. The objective function is chosen based upon some physiological imperative – in this case, to increase force sharing between the different muscles. However,



this is a numerical process and the actual activation pattern of the subject may have been quite different. In mitigation of this limitation however, we have performed quite an extensive amount of work that has demonstrated the validity and reliability of the FreeBody model (Cleather and Bull, 2015; Ding et al., 2016; Price et al., 2017) and this does allow the results of this study to be treated with some confidence.

### **The Role of the Biarticular Hamstrings in the Rotational Stability of the Tibia in the Transverse Plane**

A consideration of Figure 4 indicates that the internal/external rotation moments that each of the three biarticular hamstrings exert on the tibia are similar in magnitude to the flexion moment that they create. This in turn suggests that the creation of internal/external moments is an important aspect of their role. In addition, Figure 4 also shows that there is a co-contraction of the biarticular hamstrings, creating strong opposing internal and external rotation moments, presumably as a mechanism to create joint stability (it should be noted that the net effect is an internal rotation moment). Aside from the influence of the patellar tendon which provides a somewhat consistent internal rotation moment there are no other muscles that provide substantial internal/external rotation moments in the transverse plane on the proximal tibia, suggesting that the biarticular hamstrings play a very influential role in the active creation of transverse plane rotational stability at the knee.

There are a number of potential advantages that might arise if the biarticular hamstrings are involved in stabilising the knee. The first of these is the fact that only the proximal end of the tibia is stabilised – that is the tibia is “held” in place, while the femur is relatively free to move around the “fixed” tibia. Such an arrangement is consistent with the movement pattern exhibited during vertical jumping – the femur undergoes a considerably greater angular excursion than the tibia (Bobbert and van Soest, 2001). Similarly, during vertical jumping



subjects exhibit a characteristic proximal to distal pattern of extension (Bobbert and van Ingen Schenau, 1988; Bobbert and van Soest, 2001). That is, the femur starts to extend (i.e. to rotate back to an upright position) first, when the knees are most flexed. This corresponds to the period of peak co-contraction seen in this study, suggesting that the tibia is most strongly stabilised when the femur is extending more than the other segments.

The alternative to biarticular muscle involvement in the creation of joint stability is that instead, there is co-contraction of monoarticular muscles to stabilise the joint. However, in this case, increased co-contraction results in the articulating surfaces of the joint being directly tethered to each other increasingly strongly. This leads to increased stiffness of the joint hampering movement. The apparent paradox that increasing stability imposes a penalty on movement can be neatly resolved by characterising active “joint” stability as described in the previous paragraph. That is rather than the joint being stabilised implying that the movement of the segments relative to one another is constrained, instead one of the articulating segments is stabilised, leaving the other free to move about it.

There is an additional benefit derived from the biarticular hamstrings creating rotational stability as opposed to analogous monoarticular muscles. For instance, if the hamstrings were monoarticular flexors of the knee with a similar ability to internally and externally rotate the tibia, then co-contraction to create stability would impose a flexion moment on the femur when the musculature is trying to extend the femur. Instead, as we have shown in a previous paper (Cleather et al., 2015), the biarticular hamstrings are actually also extensors of the femur. Thus activity in the biarticular hamstrings serves a complementary purpose – stabilising the tibia while contributing to the rotation of the femur about it.

Finally, it is possible that the tibia being stabilised by muscles of the hip, simplifies the motor control of movement. That is, activation of all of the muscles of the hip to extend the femur,



simultaneously results in the tibia being stabilised. This seems to be a less vulnerable strategy for creating stability than relying on the synchronous recruitment of muscles at a distal joint. Of course, such a contention is highly speculative.

### **The Role of the Biarticular Hamstrings in Internally Rotating the Tibia During Vertical Jumping**

The net effect of the internal/external rotation moments impressed on the tibia by the biarticular hamstrings is to create internal rotation. It is notable that the peak internal rotation created by the biarticular hamstrings coincides with the peak inter-segmental internal rotation moment. Given the relatively unchanging internal/external rotation effects of the other muscles of the knee, this suggests that it is the biarticular hamstrings are responsible for this pattern.

When the hip is flexed and the foot is fixed (as in vertical jumping), abduction of the femur creates a lateral translation of the knee joint. There are two ways this can be achieved. The first option is that the lateral translation of the proximal tibia relative to the foot creates a lateral angulation of the tibial plane, creating knee varus. Alternatively, the tibia can internally rotate, which has the effect of reducing the knee varus. Consequently, an internal rotation of the tibia is arguably a functional adaptation, which improves the alignment of the knee joint.

Again, there is a potential advantage to this internal rotation being mediated by biarticular muscles. An easy way to understand this is by considering the example of gastrocnemius function in the sagittal plane during vertical jumping. In particular, if the length of gastrocnemius is held constant, then extension of the knee (increasing knee joint angle) results in the gastrocnemius pulling on the foot - plantar flexing the ankle. An analogous arrangement is present for semimembranosus and semitendinosus. That is, if their length is



constant, then when the hip is flexed, abduction of the femur in turn causes them to pull on the tibia, creating internal rotation. Thus, there is a mechanical coupling between hip abduction when the hip is flexed, and internal rotation of the tibia. Again, one can speculate that such an arrangement simplifies the motor control of movement, as the desired internal rotation of the tibia is a mechanical consequence of abduction of the femur caused simply by activation of the hip musculature rather than it requiring a separate synchronous recruitment of knee muscles.

## **Conclusions**

The results of this study support the previous work of Buford and colleagues (2001) in quantifying the internal/external rotation moment arms of the muscles of the knee. In particular, the external rotators of the tibia were biceps femoris and TFL, whereas the internal rotators were semimembranosus, semitendinosus, sartorius, gracilis, popliteus and the patellar tendon. The internal/external rotation moment arm increased as the knee flexed apart from for popliteus and the patellar tendon. The internal/external rotation moments impressed by the biarticular hamstrings on the tibia were of similar magnitude to the flexion moments they impressed on the tibia. It was argued that this provides evidence that an important role of the biarticular muscles is thus to create rotational stability of the tibia in the transverse plane, and that there are a number of advantages to the biarticular muscles performing this role.

## **DATA AVAILABILITY**

The data on which this article is based is publicly available as electronic supplementary material in Cleather and Bull (Cleather and Bull, 2015) (doi: [10.1098/rsos.140449](https://doi.org/10.1098/rsos.140449)).



## **COMPETING INTERESTS**

I have no competing interests.

## **AUTHORS' CONTRIBUTIONS**

All of the work described here was carried out by DJC.

## **FUNDING**

This study was not funded.

## **RESEARCH ETHICS**

The collection of this data set was approved the institutional review board of St Mary's University College and all subjects provided informed consent.

## **ANIMAL ETHICS**

No animals were used in this study.

## **PERMISSION TO CARRY OUT FIELDWORK**

No permission was required.



## **ACKNOWLEDGEMENTS**

I would like to thank my former student Graham Dean who first suggested to me that the biarticular muscles may play a role in the rotational stability of the knee and who thus prompted this study.



## REFERENCES

- Baratta, R., Solomonow, M., Zhou, B.H., Letson, D., Chuinard, R., Dambrosia, R., 1988. Muscular Coactivation - the Role of the Antagonist Musculature in Maintaining Knee Stability. *Am. J. Sports Med.* 16, 113–122.
- Bobbert, M.F., Huijing, P.A., van Ingen Schenau, G.J., 1986a. A model of the human triceps surae muscle-tendon complex applied to jumping. *J. Biomech.* 19, 887–898. [https://doi.org/10.1016/0021-9290\(86\)90184-3](https://doi.org/10.1016/0021-9290(86)90184-3)
- Bobbert, M.F., Huijing, P.A., van Ingen Schenau, G.J., 1986b. An estimation of power output and work done by the human triceps surae muscle-tendon complex in jumping. *J. Biomech.* 19, 899–906. [https://doi.org/10.1016/0021-9290\(86\)90185-5](https://doi.org/10.1016/0021-9290(86)90185-5)
- Bobbert, M.F., van Ingen Schenau, G.J., 1988. Coordination in vertical jumping. *J. Biomech.* 21, 249–262. [https://doi.org/10.1016/0021-9290\(88\)90175-3](https://doi.org/10.1016/0021-9290(88)90175-3)
- Bobbert, M.F., van Soest, A.J., 2001. Why do people jump the way they do? *Exerc. Sport Sci. Rev.* 29, 95–102.
- Buford Jr, W.L., Ivey Jr, F.M., Nakamura, T., Patterson, R.M., Nguyen, D.K., 2001. Internal/external rotation moment arms of muscles at the knee: moment arms for the normal knee and the ACL-deficient knee. *The Knee* 8, 293–303. [https://doi.org/10.1016/S0968-0160\(01\)00106-5](https://doi.org/10.1016/S0968-0160(01)00106-5)
- Cleather, D.J., Bull, A.M.J., 2015. The development of a segment-based musculoskeletal model of the lower limb: introducing FreeBody. *R. Soc. Open Sci.* 2, 140449.
- Cleather, D.J., Bull, A.M.J., 2012. The development of lower limb musculoskeletal models with clinical relevance is dependent upon the fidelity of the mathematical description of the lower limb. Part 1: equations of motion. *Proc. Inst. Mech. Eng. [H]* 226, 120–132.
- Cleather, D.J., Bull, A.M.J., 2011a. An Optimization-Based Simultaneous Approach to the Determination of Muscular, Ligamentous, and Joint Contact Forces Provides Insight into Musculoligamentous Interaction. *Ann. Biomed. Eng.* 39, 1925–1934. <https://doi.org/10.1007/s10439-011-0303-8>
- Cleather, D.J., Bull, A.M.J., 2011b. Knee and hip joint forces: Sensitivity to the degrees of freedom classification at the knee. *Proc. Inst. Mech. Eng. [H]* 225, 621–626.
- Cleather, D.J., Bull, A.M.J., 2010a. Lower extremity musculoskeletal geometry effects the calculation of patellofemoral forces in vertical jumping and weightlifting. *Proc. Inst. Mech. Eng. Part H - J. Eng. Med.* 224, 1073–1083.
- Cleather, D.J., Bull, A.M.J., 2010b. Influence of inverse dynamics methods on the calculation of inter-segmental moments in vertical jumping and weightlifting. *Biomed. Eng. OnLine* 9, 74. <https://doi.org/10.1186/1475-925X-9-74>
- Cleather, D.J., Goodwin, J.E., Bull, A.M.J., 2011a. An Optimization Approach to Inverse Dynamics Provides Insight as to the Function of the Biarticular Muscles During Vertical Jumping. *Ann. Biomed. Eng.* 39, 147–160. <https://doi.org/10.1007/s10439-010-0161-9>
- Cleather, D.J., Goodwin, J.E., Bull, A.M.J., 2011b. Erratum to: An Optimization Approach to Inverse Dynamics Provides Insight as to the Function of the Biarticular Muscles During Vertical Jumping. *Ann. Biomed. Eng.* 39, 2476–2478. <https://doi.org/10.1007/s10439-011-0340-3>
- Cleather, D.J., Southgate, D.F.L., Bull, A.M.J., 2015. The role of the biarticular hamstrings and gastrocnemius muscles in closed chain lower limb extension. *J. Theor. Biol.* 365, 217–225. <https://doi.org/10.1016/j.jtbi.2014.10.020>



- Cleather, D.J., Southgate, D.F.L., Bull, A.M.J., 2014. On the role of the patella, ACL and joint contact forces in the extension of the knee. *PloS One* 9, e115670.  
<https://doi.org/10.1371/journal.pone.0115670>
- Czasche, M.B., Goodwin, J.E., Bull, A.M.J., Cleather, D.J., 2017. The effects of an eight week strength training intervention on tibiofemoral joint loading during landing: A cohort study. *BMJ Open Sport Exerc. Med.* 3, e000273.  
<https://doi.org/10.1136/bmjsem-2017-000273>
- de Leva, P., 1996. Adjustments to Zatsiorsky - Seluyanov's segment inertia parameters. *J. Biomech.* 29, 1223–1230.
- DeFrate, L.E., Gill, T.J., Li, G., 2004. In Vivo Function of the Posterior Cruciate Ligament During Weightbearing Knee Flexion. *Am. J. Sports Med.* 32, 1923–1928.  
<https://doi.org/10.1177/0363546504264896>
- Ding, Z., Nolte, D., Kit Tsang, C., Cleather, D.J., Kedgley, A.E., Bull, A.M.J., 2016. In Vivo Knee Contact Force Prediction Using Patient-Specific Musculoskeletal Geometry in a Segment-Based Computational Model. *J. Biomech. Eng.* 138, 021018–021018.  
<https://doi.org/10.1115/1.4032412>
- Dumas, R., Aissaoui, R., de Guise, J.A., 2004. A 3D generic inverse dynamic method using wrench notation and quaternion algebra. *Comput. Methods Biomech. Biomed. Engin.* 7, 159–166.
- Klein Horsman, M.D., Koopman, H.F.J.M., van der Helm, F.C.T., Poliacu Prose, L., Veeger, H.E.J., 2007. Morphological muscle and joint parameters for musculoskeletal modelling of the lower extremity. *Clin. Biomech.* 22, 239–247.  
<https://doi.org/10.1016/j.clinbiomech.2006.10.003>
- Parr, M., Price, P.D., Cleather, D.J., 2017. Effect of a gluteal activation warm-up on explosive exercise performance. *BMJ Open Sport Exerc. Med.* 3, e000245.  
<https://doi.org/10.1136/bmjsem-2017-000245>
- Price, P.D.B., Gissane, C., Cleather, D.J., 2017. Reliability and Minimal Detectable Change Values for Predictions of Knee Forces during Gait and Stair Ascent Derived from the FreeBody Musculoskeletal Model of the Lower Limb. *Front. Bioeng. Biotechnol.* 5.  
<https://doi.org/10.3389/fbioe.2017.00074>
- Price, P.D.B., Gissane, C., Cleather, D.J., 2016. The evaluation of the FreeBody lower limb model during activities of daily living. Presented at the Bioengineering16.  
<https://doi.org/10.13140/RG.2.2.29146.34241>
- Southgate, D.F., Cleather, D.J., Weinert-Aplin, R.A., Bull, A.M., 2012. The sensitivity of a lower limb model to axial rotation offsets and muscle bounds at the knee. *Proc. Inst. Mech. Eng. [H]* 226, 660–669. <https://doi.org/10.1177/0954411912439284>
- Van Ingen Schenau, G., Bobbert, M., 1993. The global design of the hindlimb in quadrupeds. *Acta Anat. (Basel)* 146, 103–108.