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2 **Hip and knee joint loading during vertical jumping and push jerking**

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18

19 **Abstract**

20 *Background*

21 The internal joint contact forces experienced at the lower limb have been frequently studied
22 in activities of daily living and rehabilitation activities. In contrast, the forces experienced
23 during more dynamic activities are not well understood, and those studies that do exist
24 suggest very high degrees of joint loading.

25 *Methods*

26 In this study a biomechanical model of the right lower limb was used to calculate the internal
27 joint forces experienced by the lower limb during vertical jumping, landing and push jerking
28 (an explosive exercise derived from the sport of Olympic weightlifting), with a particular
29 emphasis on the forces experienced by the knee.

30 *Findings*

31 The knee experienced mean peak loadings of $2.4-4.6 \times$ body weight at the patellofemoral
32 joint, $6.9-9.0 \times$ body weight at the tibiofemoral joint, $0.3-1.4 \times$ body weight anterior tibial
33 shear and $1.0-3.1 \times$ body weight posterior tibial shear. The hip experienced a mean peak
34 loading of $5.5-8.4 \times$ body weight and the ankle $8.9-10.0 \times$ body weight.

35 *Interpretation*

36 The magnitudes of the total (resultant) joint contact forces at the patellofemoral joint,
37 tibiofemoral joint and hip are greater than those reported in activities of daily living and less
38 dynamic rehabilitation exercises. The information in this study is of importance for medical
39 professionals, coaches and biomedical researchers in improving the understanding of acute
40 and chronic injuries, understanding the performance of prosthetic implants and materials,
41 evaluating the appropriateness of jumping and weightlifting for patient populations and
42 informing the training programmes of healthy populations.

Hip and knee joint loading

43

44 **Keywords:** musculoskeletal modelling; inverse dynamics; joint contact forces; vertical

45 jumping; weightlifting

46

47 **Introduction**

48 The quantification of the forces experienced by the hip and knee during movement has been
49 of great interest to the biomedical research community and there have been a large number of
50 studies that have sought to quantify this loading through both musculoskeletal modelling
51 techniques and direct measurement. The majority of these studies have focussed on activities
52 of daily living (ADLs; movements like “sit to stand”, “stand to sit”, gait, stair ascent/descent),
53 or rehabilitation exercises characterized by relatively slow execution speeds (exercises like
54 the squat or lunge). The breadth of this literature, allows a typical, albeit quite wide, range
55 for the loading during these types of activities to be suggested. For instance, at least 15
56 different groups have calculated internal knee forces during squatting using musculoskeletal
57 modelling techniques (Collins, 1994; Dahlkvist et al., 1982; Escamilla et al., 1998; Nagura et
58 al., 2006; Nisell, 1985; Reilly and Martens, 1972; Salem and Powers, 2001; Sharma et al.,
59 2008; Shelburne and Pandy, 1998; Shelburne and Pandy, 2002; Smith et al., 2008;
60 Thambyah, 2008; Toutoungi et al., 2000; Wallace et al., 2002; Wilk et al., 1996) and the
61 internal forces suggested during body weight squatting include a patellofemoral joint force
62 (PFJF) range of $2.5-7.6 \times BW$ and a tibiofemoral joint force (TFJF) range of $2.5-7.3 \times BW$.

63 In contrast, there are fewer musculoskeletal modelling studies that have sought to understand
64 the loading of the hip and knee joints during more dynamic movements with faster execution
65 speeds. Those studies that do exist are often based on simple biomechanical models with
66 inherently limiting assumptions and which thus may not accurately capture the nature of the
67 joint loading (Nisell and Mizrahi, 1988; Simpson and Kanter, 1997; Simpson and Pettit,
68 1997; Smith, 1975). In particular, there is a tendency for these studies to report joint loadings

69 that seem very high in comparison to those found in ADLs, even when accounting for a
70 premium attributable to the more demanding nature of these activities. For example,
71 Simpson and colleagues (Simpson et al., 1996; Simpson and Kanter, 1997; Simpson and
72 Pettit, 1997) found that during a landing from a travelling jump (a horizontal jump to a single
73 leg landing) the PFJF was $10.4 \times BW$ and the TFJF $16.8 \times BW$. Similarly, in a pioneering
74 study, Smith (1975) suggested that the TFJF experienced during a jump landing was in the
75 range of $17.0-24.4 \times BW$. These high values may be a result of the lack of detail in the
76 biomechanical models employed (Cleather and Bull, 2010b; Cleather and Bull, 2012b) or
77 even inaccurate model assumptions. In recent years, the prevalence of sporting injuries to the
78 anterior cruciate ligament (ACL) of the knee (Majewski et al., 2006) has prompted an interest
79 in quantifying the loading of this structure during movement, also by employing
80 musculoskeletal modelling techniques (Kernozek and Ragan, 2008; Pflum et al., 2004). A
81 common approach is to calculate the anterior shear force (that is the force that tends to
82 displace the tibia anteriorly on femur) and to use this as a proxy for the ACL loading (as the
83 ACL is the primary restraint to anterior drawer of the knee). However, these studies also tend
84 to be based upon inappropriately simple biomechanical models (Sell et al., 2007; Yu et al.,
85 2006), and thus even a clear idea as to the shear forces experienced by the knee is largely
86 unknown.

87 The development of instrumented prostheses has permitted the *in vivo* measurement of forces
88 in the hip and knee, and provided new insights. For instance, D'Lima and colleagues have
89 shown that during ADLs the magnitude of the TFJF is in the range of $2.0-3.0 \times BW$, but that
90 during sporting activities (including jogging, tennis and golf) this rises to $3.0-4.5 \times BW$
91 (D'Lima et al., 2005b; D'Lima et al., 2005a; D'Lima et al., 2006; D'Lima et al., 2007; D'Lima
92 et al., 2008). These values also seem to suggest that the higher internal forces predicted

93 during more dynamic activities by earlier biomechanical models could be questionable. The
94 highly invasive nature of this research restricts these studies to patient populations (of often
95 advanced ages) however, and it does seem likely that young, healthy populations might
96 experience a greater loading.

97 It is clear that the magnitude of the forces experienced by the hip and knee joints during
98 dynamic activities characterized by rapid movement speeds is not well understood. In
99 particular, a typical upper range for the loading of the hip and knee joints in sporting
100 movements in young healthy populations is generally unknown. The purpose of this study
101 was therefore to use a previously developed model of the musculoskeletal model of the lower
102 limb (Cleather et al., 2011a; Cleather et al., 2011b; Cleather and Bull, 2010b) to quantify the
103 nature and magnitude of the forces experienced at the joints of the lower extremity by a
104 young athletic male population during vertical jumping and push jerking (two lower
105 extremity activities characterized by high movement speeds and force loading and that are
106 similar in kinematic character) with a particular focus on the forces experienced by the knee.

107 **Methods**

108 In this study a previously described biomechanical model (Cleather, 2010; Cleather et al.,
109 2011a; Cleather et al., 2011b; Cleather and Bull, 2010a; Cleather and Bull, 2010b) of the
110 right lower limb was employed to calculate the internal joint forces produced during vertical
111 jumping and push jerking. The validation and verification of the model has been described in
112 previous work (Cleather, 2010; Cleather et al., 2011a; Cleather et al., 2011b; Cleather and
113 Bull, 2010b) as has the sensitivity of the model to some key parameters (Cleather, 2010;

114 Cleather and Bull, 2010a; Cleather and Bull, 2010b; Cleather and Bull, 2011). The study was
115 approved by the local research ethics committee and all participants provided informed
116 consent. Twelve athletic males (mean age 27.1 SD 4.3 years; mean mass 83.7 SD 9.9 kg)
117 were recruited to take part in this study. After performing a standardized warm up consisting
118 of lower extremity body weight exercises (such as squats, lunges and vertical jumps) each
119 subject performed 5 maximal countermovement jumps with their hands on their hips and the
120 highest jump (mean height 0.38 SD 0.05 m) was chosen for analysis. Nine of the subjects
121 (mean age 27.3 SD 4.1 years; mean mass 84.1 SD 10.7 kg) who were familiar with the push
122 jerk exercise (more than six months experience in Olympic weightlifting) also performed 3
123 repetitions of a push jerk with 40 kg – a movement derived from the competitive sport of
124 Olympic weightlifting where a barbell is thrust overhead primarily by forces produced by
125 extension of the lower limb joints. The data set comprised the position of reflective markers
126 placed on key anatomical landmarks (Van Sint Jan, 2005; Van Sint Jan and Croce, 2005)
127 determined using the Vicon motion capture system (Vicon MX System, Vicon Motion
128 Systems Ltd, Oxford, UK) and the ground reaction force recorded by a portable force plate
129 (Kistler Type 9286AA, Kistler Instrumente AG, Winterthur, Switzerland). The marker set
130 employed in this study is described in detail elsewhere (Cleather, 2010), and comprises
131 markers on the pelvis (4 markers on the anterior and posterior supra-iliac spines), thigh (5
132 markers – including markers on the medial and lateral epicondyles), calf (5 markers –
133 including markers on the medial and lateral epicondyles) and foot (4 markers – including
134 markers on the rear of calcaneus and the head of the second metatarsal). As the
135 musculoskeletal model is of the right limb alone, each subject performed each trial with only
136 their right foot on the force plate, thus the ground reaction force was that impressed by the
137 right limb alone. All data was collected at 200 Hz. The raw data was filtered using
138 generalized cross validatory spline filtering (Woltring, 1986; otherwise known as a Woltring

139 filter) using a 5th order spline and a cut-off frequency of 10 Hz. Following the
140 recommendation of Bisseling and Hof (2006), the force data was filtered using the same cut-
141 off frequency as the kinematic data.

142 The musculoskeletal model consists of a linked series of four segments representing the foot,
143 calf, thigh and pelvis articulated by ball and socket joints at the ankle, knee and hip. After
144 filtering these segments were constructed from the positions of the markers using the method
145 of Horn (1987) to establish the position and orientation of each segment. The anthropometry
146 used in the model was taken from the work of de Leva (1996).

147 The data of Klein Horsman and colleagues (2007) was used to create a subject-specific
148 musculoskeletal geometry of the lower limb. This consisted of 163 different line elements
149 representing 38 different muscles of the lower limb. The position of the patella relative to the
150 femur was calculated using the Klein Horsman data to determine the position of the patellar
151 origin relative to the femur as a function of the knee flexion angle. The orientation of the
152 patella relative to the femur (i.e. its sagittal plane rotation) for a given knee flexion angle was
153 calculated using the data of Nha and colleagues (2008) using spline interpolation (using
154 "Numerical Recipes in C++"; Press et al., 2002). The patellofemoral joint model also
155 included the addition of via points to model the wrapping of the quadriceps around the
156 femoral condyles in deep knee flexion. This was achieved by simply defining a via point for
157 each quadriceps muscle element through which the element was constrained to pass once the
158 quadriceps had begun to wrap around the femoral condyles. Finally, the changing ratio
159 between quadriceps and patellar tendon forces (Mason et al., 2008) with increased knee
160 flexion angle was calculated based upon the geometrical relationship between patella, patellar

161 tendon and quadriceps tendons assuming the maintenance of force and moment equilibrium
 162 at the patella.

163 Muscle forces were determined using an optimization based approach to inverse dynamics
 164 (Cleather, 2010; Cleather et al., 2011a; Cleather et al., 2011b). The inverse dynamics method
 165 of Dumas and colleagues (2004) was used to formulate the equations of motion of each
 166 segment as a function of the unknown muscle forces:

$$\begin{aligned}
 167 \quad & \left[\begin{array}{c} \hat{S}_i \\ \sum_{j=1}^K F_j \cdot \hat{r}_{ji} \times \hat{n}_{ji} - \sum_{j=1}^K F_j \cdot \hat{r}_{j(i-1)} \times \hat{n}_{j(i-1)} + \sum_{j=1}^K b_{ji} F_j \cdot \hat{d}_i \times \hat{n}_{j(i-1)} \end{array} \right] = \\
 & \begin{bmatrix} m_i E_{3 \times 3} & 0_{3 \times 3} \\ m_i \tilde{c}_i & I_i \end{bmatrix} \begin{bmatrix} \hat{a}_i - \hat{g} \\ \ddot{\theta}_i \end{bmatrix} + \begin{bmatrix} 0_{3 \times 1} \\ \dot{\theta}_i \times I_i \dot{\theta}_i \end{bmatrix} + \begin{bmatrix} E_{3 \times 3} & 0_{3 \times 3} \\ \tilde{d}_i & E_{3 \times 3} \end{bmatrix} \begin{bmatrix} \hat{S}_{i-1} \\ \hat{M}_{i-1} \end{bmatrix}
 \end{aligned}$$

168 $b_{ji} = 1$ for biarticular muscles that cross but do not attach to segment i ;

169 $b_{ji} = 0$ for all other muscles (1)

170 Where \hat{M}_{i-1} was set to zero for $i > 1$ and:

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i	– segment number or joint number (1 represents the most distal segment or joint)
\hat{S}_i	– proximal joint reaction forces
\hat{S}_{i-1}	– distal joint reaction forces
\hat{M}_{i-1}	– distal joint moments
I_i	– inertia tensor
$\dot{\hat{\theta}}_i$	– angular velocity about COM
$\ddot{\hat{\theta}}_i$	– angular acceleration about COM
m_i	- segment mass
$E_{3 \times 3}$	- identity matrix
\hat{a}_i	- linear acceleration of segment COM
\hat{c}_i	- vector from the proximal joint to the segment COM
\hat{d}_i	- vector from the proximal to the distal joint
\hat{g}	- acceleration due to gravity
K	– number of muscles
F_j	– individual muscle force
F_{\max_j}	– maximum possible muscle force
\hat{n}_{ji}	– line of action of muscle j about joint i
\hat{r}_{ji}	– moment arm of muscle j about joint i

171 and \tilde{c} and \tilde{d} represent the skew symmetric matrix of a 3D vector:

$$172 \quad \tilde{c} = \begin{pmatrix} 0 & -c_3 & c_2 \\ c_3 & 0 & -c_1 \\ -c_2 & c_1 & 0 \end{pmatrix} \quad (2)$$

173 Equation 1 represents a system of 9 equations of inter-segmental force equilibrium that are
174 determinate and 9 equations of moment equilibrium that are indeterminate (the equations of
175 motion are posed by considering the foot, calf and thigh segments). The indeterminate
176 problem is solved with an optimization approach by seeking to minimize the cost function of
177 Crowninshield and Brand (1981). This cost function is based upon minimizing the sum of
178 the muscle stress raised to the power n . It has previously been shown that this solution tends
179 towards a limit with increasing n (Rasmussen et al., 2001), and in this model a value of $n =$
180 30 produces physiologically realistic results (Cleather et al., 2011a):

$$181 \quad \min_{F_j} f = \sum_{i=j}^{163} \left(\frac{F_j}{F_{\max_j}} \right)^{30} \quad (3)$$

182 Subject to the constraints that:

$$183 \quad 0 \leq F_j \leq F_{\max_j} \quad (4)$$

184 The maximum muscle force was calculated by multiplying the physiological cross-sectional
185 area of each muscle given in the Klein Horsman data set (2007) by an assumed maximum
186 muscle stress ($3.139 \times 10^5 \text{ N/m}^2$; Yamaguchi, 2001).

187 Finally, the calculated muscle forces were combined with the inter-segmental forces to
188 calculate internal joint forces. The magnitude of the total joint reaction force was calculated
189 for the ankle (AF), patellofemoral joint (PFJF), tibiofemoral joint (TFJF) and the hip (HF).
190 In addition, the anterior and posterior shear at the tibiofemoral joint (AS and PS; presented in
191 the tibial coordinate frame) was also computed in an effort to understand the loading
192 experienced by the cruciate ligaments of the knee. A repeated measures ANOVA with post
193 hoc Bonferroni corrected pair wise comparisons was used to evaluate whether the joint forces
194 experienced in each activity were different. A significance level of $p < 0.05$ was set a priori.

195 **Results**

196 The optimization found a solution for over 99% of the frames of interest. Where a solution
197 could not be found the frame was omitted from the results. Table 1 presents the mean peak
198 forces in the lower limb during jump takeoff, landing, the push jerk drive and the push jerk
199 catch. There were significant differences in the forces experienced at the PFJ (jumping
200 significantly greater than jerk catching and jerk drive significantly greater than jerk catching
201 – $p < 0.05$) and in posterior shear at the TFJ (again, both jumping and jerking significantly
202 greater than jerk catching – $p < 0.05$). The ankle joint experienced the greatest loading
203 whereas the PFJ was loaded the least during all activities.

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204 Table 1. Mean (SD) peak normalized forces (\times BW) calculated during the four activities (* =
 205 $p < 0.05$, when compared to jumping; ‡ = $p < 0.05$, when compared to jerking; # = $p < 0.05$,
 206 when compared to catching).

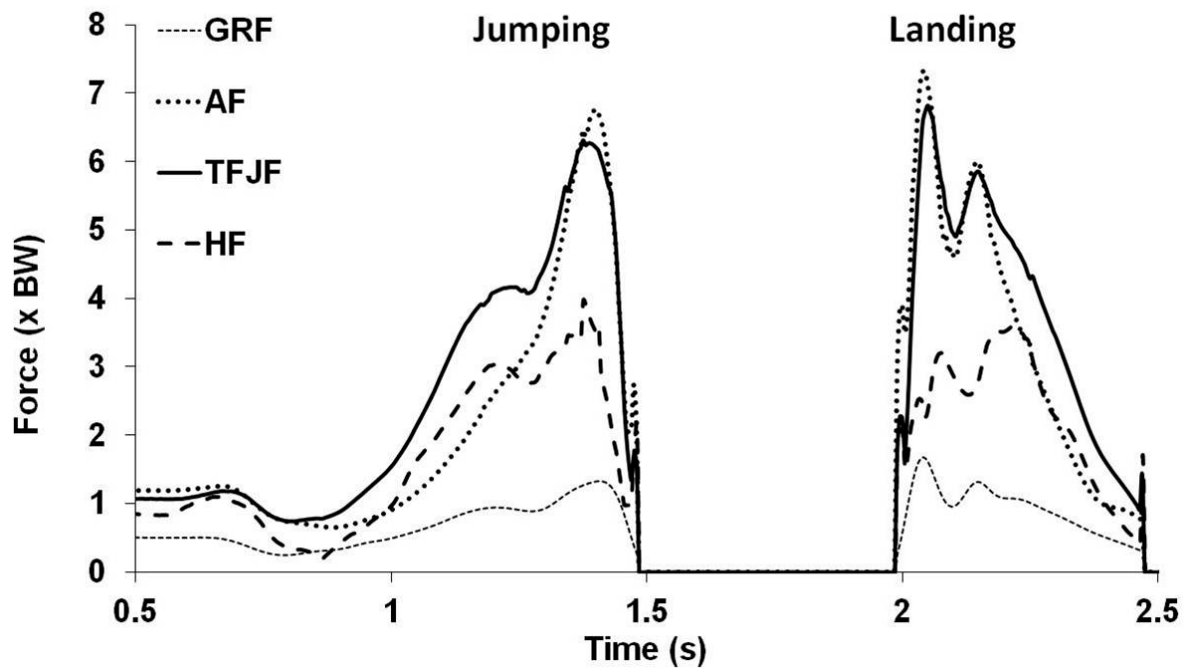
	Ankle	Knee				Hip	GRF
		PFJ	TFJ	AS	PS		
Jump n=12	8.9 (1.8)	4.2 (1.2) #	6.9 (1.0)	0.6 (0.4)	2.4 (1.1) #	5.5 (1.1)	1.3 (0.1) ‡
Land n=12	9.7 (4.1)	3.6 (0.9)	7.6 (2.1)	0.7 (0.5)	2.0 (1.2)	6.0 (3.0)	1.6 (0.3)
Jerk n=9	10.0 (1.4)	4.6 (1.2) #	9.0 (1.3)	0.3 (0.2)	3.1 (1.6) #	7.8 (4.2)	1.6 (0.2) *#
Catch n=9	9.9 (5.6)	2.4 (0.9) *‡	7.8 (4.0)	1.4 (1.4)	1.0 (0.9) *‡	8.4 (7.8)	1.1 (0.2) ‡

207

208 Figures 1 and 2 present the forces experienced at the knee by a typical subject during vertical
 209 jumping and landing. In particular, Figure 2 illustrates the shear loading of the knee where
 210 positive shear represents posterior tibial shear (i.e. the posterior cruciate ligament (PCL) is
 211 loaded), whereas anterior tibial shear (loading of the ACL) is negative. The majority of
 212 subjects experienced a consistent pattern in the shear loading of the knee during takeoff.
 213 Early in the takeoff there was a small degree of anterior shear, which was followed by a
 214 sustained posterior shear for the bulk of the jump. Finally, immediately prior to takeoff the

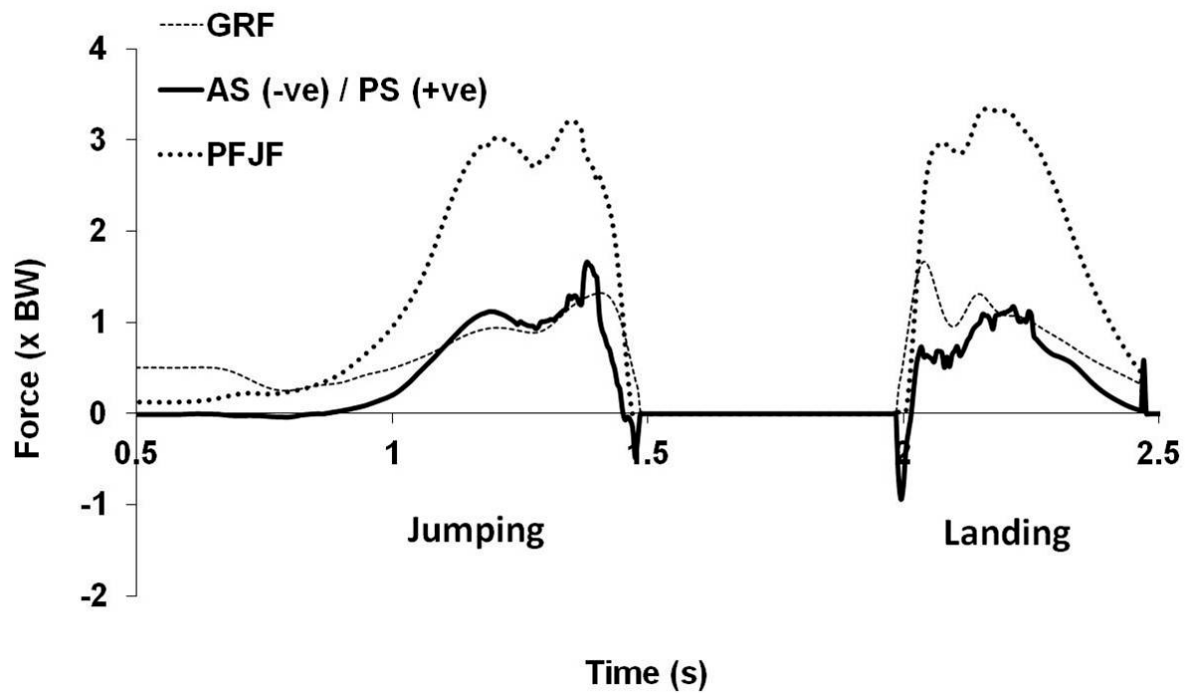
215 shear force oscillated between anterior and posterior. There was more variation in the pattern
216 of shear loading during landing, although there were some weak trends. Initial impact was
217 generally associated with at least one spike directed anteriorly (in many trials there was a
218 period of variability, which sometimes resulted in the direction of shear loading switching
219 multiple times between anterior and posterior) before a more sustained period of posterior
220 shearing.

221 Figure 1. Ankle, TFJF and hip loading experienced by a typical subject during vertical
222 jumping and landing (AF = ankle joint reaction force; TFJF = tibiofemoral joint reaction
223 force; HF = hip joint reaction force; GRF = ground reaction
224 force).



225

226 Figure 2. PFJF and tibial shear experienced by a typical subject during vertical jumping and
227 landing (PFJF = patellofemoral joint reaction force; AS = anterior shear; PS = posterior
228 shear; GRF = ground reaction force).



229

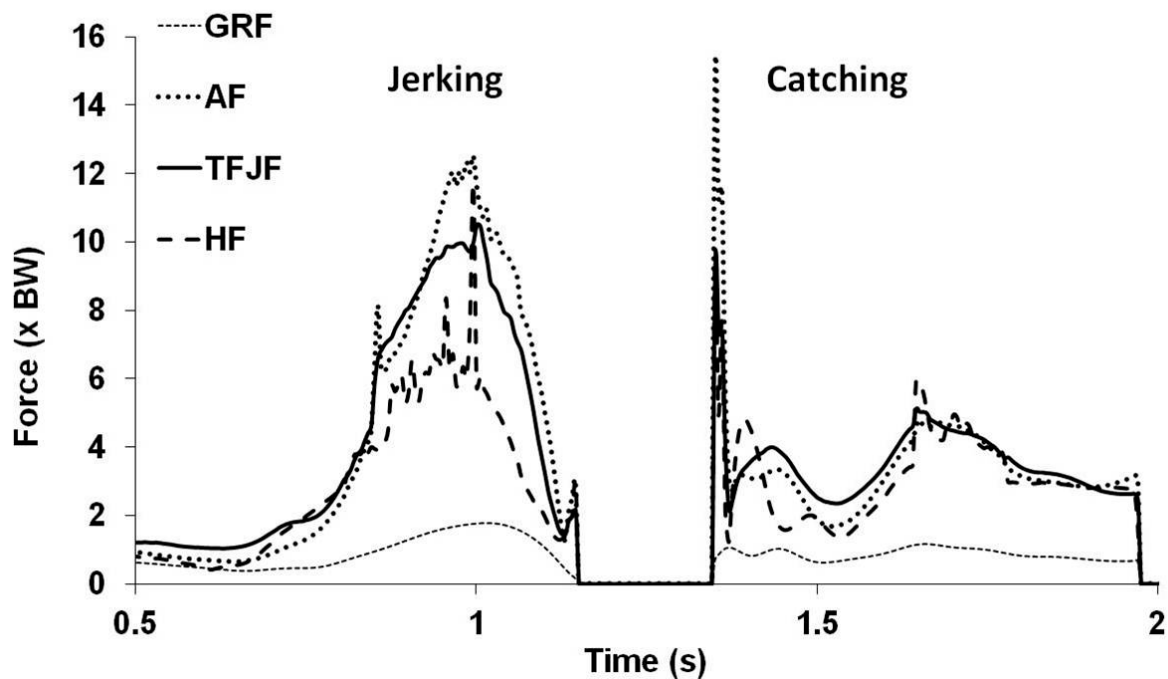
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232

233 Figures 3 and 4 present the forces experienced at the knee by a typical subject during push
234 jerking. During the push jerk drive for the majority of subjects the pattern of tibial shear was
235 similar to that experienced during jumping with principally posterior shearing followed by
236 brief anterior shearing. In contrast to the jump landing, no clear pattern in the direction or
237 loading of tibial shear during push jerk catching emerged (although there was a weak trend
238 towards early brief anterior shear followed by sustained posterior shearing).

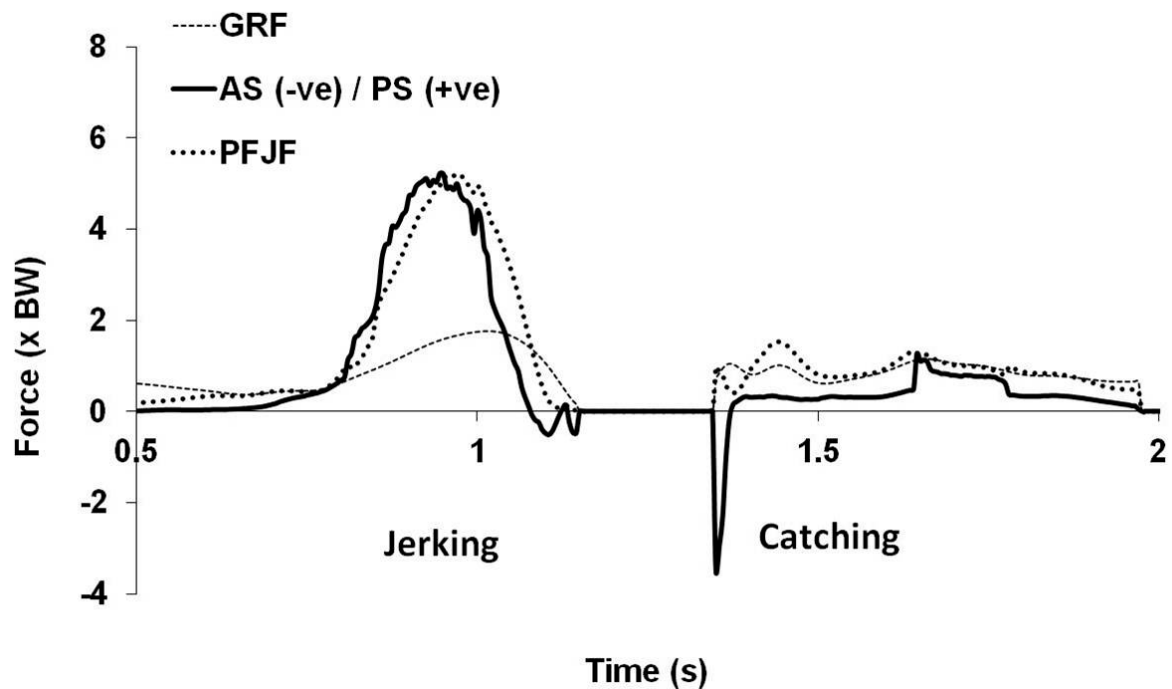
239 Figure 3. Ankle, TFJF and hip loading experienced by a typical subject during the jerking
240 and catching (AF = ankle joint reaction force; TFJF = tibiofemoral joint reaction force; HF =
241 hip joint reaction force; GRF = ground reaction force).



242

243

244 Figure 4. PFJF and tibial shear experienced by a typical subject during jerking and catching
245 (PFJF = patellofemoral joint reaction force; AS = anterior shear; PS = posterior shear; GRF =
246 ground reaction force).



247

248 Discussion

249 In this study a previously described musculoskeletal model of the right lower limb was used
250 to calculate the internal joint forces experienced during vertical jumping, landing and jerking
251 with a focus on the knee. In general, the forces experienced at each joint were of similar
252 magnitude in all four activities, although there were some statistically significant differences.
253 When the data is taken as a whole, the TFJ appeared to experience a peak loading in the
254 range of $6.9-9.0 \times BW$, the ankle joint a loading of $8.9-10.0 \times BW$ and the hip joint a loading

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255 of $5.5-8.4 \times BW$. The peak PFJF was in the range of $2.4-4.6 \times BW$ and the AS at the tibia
256 were in the range $0.3-1.4 \times BW$ whereas the PS was of the order of $1.0-3.1 \times BW$.

257 The magnitude of the total internal hip (HF) and knee (PFJF and TFJF) forces found in this
258 work are greater than those that have been suggested to occur during ADLs and rehabilitation
259 exercises characterized by slow movement speeds, or than have been measured in patient
260 populations (Bergmann et al., 2001; D'Lima et al., 2005a; D'Lima et al., 2006; D'Lima et al.,
261 2007; D'Lima et al., 2008; Escamilla et al., 2008a; Escamilla et al., 2008b; Escamilla et al.,
262 2009a; Escamilla et al., 2009b). For instance, D'Lima and colleagues (2008) have reported
263 that the highest tibial loadings recorded by a telemeterized knee implant in a patient
264 population were between 3.0 and $4.5 \times BW$ during jogging, golf and tennis. This is a finding
265 that might be expected given that the activities considered in this study are typified by faster
266 segmental accelerations and higher ground reaction forces, and supported by the fact that
267 previous research has suggested that activities like vertical jumping and weightlifting yield
268 greater joint reaction forces (Collins, 1994; Simpson et al., 1996; Simpson and Kanter, 1997;
269 Simpson and Pettit, 1997). Despite this, the total knee joint contact forces (PFJF and TFJF
270 only) suggested by this study are less than half as great as those suggested by previous
271 analyses of vertical jumping (Simpson et al., 1996; Simpson and Kanter, 1997; Simpson and
272 Pettit, 1997; Smith, 1975).

273 Previous research that has evaluated the tensile strength of the cruciate ligaments suggests a
274 failure limit of around 2 kN for the ACL of young healthy males (Chandrashekar et al., 2006;
275 Noyes and Grood, 1976; Woo et al., 1991) and 4.5 kN for the PCL (Amis et al., 2003). This
276 study suggests a mean peak anterior shear during jumping, landing and jerking in the range of

277 240-1150 N and a mean peak posterior shear of 820-2550 N. The cruciate ligaments provide
278 the primary restraint to anterior-posterior shear at the knee joint and these values are well
279 within the ranges that could potentially be borne by the cruciate ligaments.

280 There is a growing body of evidence that asserts the importance of modelling the
281 musculoskeletal system with appropriate detail as to provide physiologically realistic results
282 (Cleather and Bull, 2012a; Cleather and Bull, 2012b). Previous studies as to the internal knee
283 forces experienced during vertical jumping have been limited by the simplifying assumptions
284 employed in order to permit a solution. These have included a lack of detail (in terms of the
285 number and variability of force actuators; Cleather and Bull, 2010b; Valente et al., 2012) or
286 the employment of only 2D models. The strength of the current work is that it is based upon
287 a well posed model that is 3D and incorporates more detail than previous studies of these
288 activities. Despite this the model lacked an adequate number of force actuators to easily find
289 a solution for a limited number of frames immediately before take-off or after landing for
290 some subjects (a higher force upper bound for the muscles was required to find a solution).
291 This may suggest that when considering these types of activities an even greater degree of
292 subject-specific detail may be necessary. An interesting question is whether the difficulty in
293 finding a solution is representative of a physiological imperative (in which case the relatively
294 higher joint forces found in these cases may be representative of the true loading) or whether
295 it is an artefact of a less well posed approximation of the geometry (in which case the
296 calculated force is likely to be an upper bound for the loading). It should also be noted that if
297 the musculoskeletal model has a more favourable geometry than the actual subject then the
298 model may underestimate the joint loading (Southgate et al., 2012), which further illustrates
299 the importance of future work to understand the effect of changes in subject-specific detail on
300 this type of model.

301 Other potential limitations of the model include the use of a cost function that is predicated
302 upon the imperative to maximize muscular endurance (and thus may not represent the motor
303 control strategy employed during maximal vertical jumping). Equally, the model has a lack
304 of detail describing the tibiofemoral joint (which does not separate the loading experienced
305 by the lateral and medial compartments).

306 An understanding as to the forces experienced by the hip and knee is of critical importance
307 for a variety of medical professionals, coaches and biomedical researchers. The importance
308 of the current study is therefore in defining a range for the joint contact forces that may be
309 experienced by athletic subjects during routine sporting activities. This study suggests that
310 the total joint contact forces experienced at the knee and hip during vertical jumping and push
311 jerking are larger than in ADLs or slower rehabilitation exercises, but that forces at the knee
312 are smaller than had been indicated in previous studies.

313 **Acknowledgements**

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315 considered in this study.

316

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