Hip and knee joint loading during vertical jumping and push jerking.

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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 × body weight at the patellofemoral

32 joint, 6.9-9.0 × body weight at the tibiofemoral joint, 0.3-1.4 × body weight anterior tibial

33 shear and 1.0-3.1 × body weight posterior tibial shear. The hip experienced a mean peak

34 loading of 5.5-8.4 × body weight and the ankle 8.9-10.0 × 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.

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 × BW and a tibiofemoral joint force (TFJF) range of 2.5-7.3 × 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 $17.0\text{-}24.4 \times$ 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 $2.0\text{-}3.0 \times$ BW, but that

90   during sporting activities (including jogging, tennis and golf) this rises to $3.0\text{-}4.5 \times$ 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 where 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 epicdonyles) 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:

167
$$
\begin{bmatrix} \hat{S}_i \\ \sum_{j=1}^{K} F_j \cdot \hat{r}_{ji} \times \hat{n}_{ji} - \sum_{j=1}^{K} F_j \cdot \hat{r}_{j(i-1)} \times \hat{n}_{j(i-1)} + \sum_{j=1}^{K} b_{ji} F_j \cdot \hat{d}_i \times \hat{n}_{j(i-1)} \end{bmatrix} =
$$
$$
\begin{bmatrix} m_i E_{3\times3} & 0_{3\times3} \\ m_i \tilde{c}_i & I_i \end{bmatrix} \begin{bmatrix} \hat{a}_i - \hat{g} \\ \ddot{\hat{\theta}}_i \end{bmatrix} + \begin{bmatrix} 0_{3\times1} \\ \dot{\theta}_i \times I_i \dot{\hat{\theta}}_i \end{bmatrix} + \begin{bmatrix} E_{3\times3} & 0_{3\times3} \\ \tilde{d}_i & E_{3\times3} \end{bmatrix} \begin{bmatrix} \hat{S}_{i-1} \\ \hat{M}_{i-1} \end{bmatrix}
$$

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:

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_{3x3}	- 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
$$\tilde{c} = \begin{pmatrix} 0 & -c_3 & c_2 \\ c_3 & 0 & -c_1 \\ -c_2 & c_1 & 0 \end{pmatrix} \qquad (2)$$

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

182   Subject to the constraints that:

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

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 (3.139×10^{5} 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.

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).
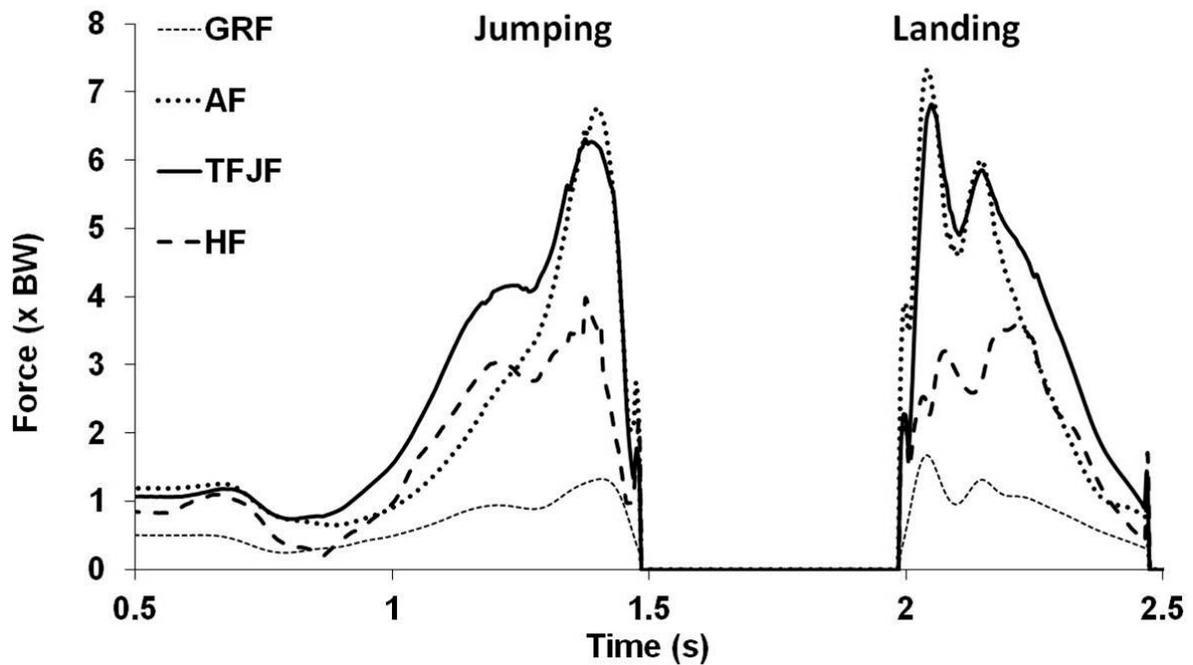
225

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
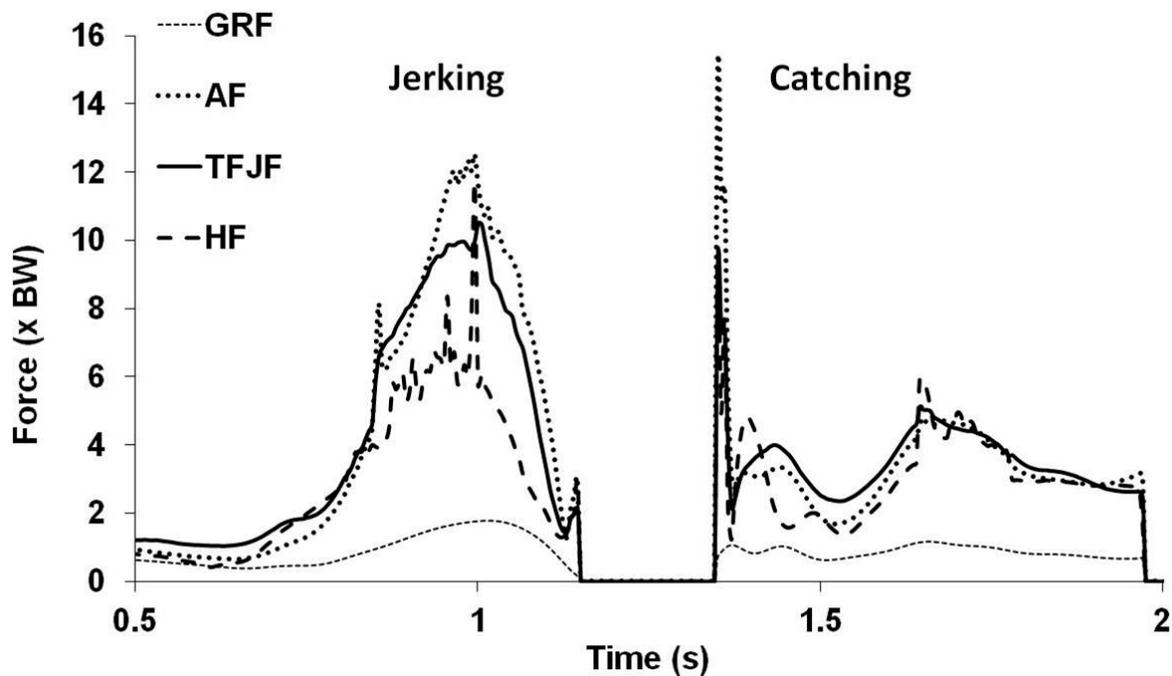
228   shear; GRF = ground reaction force).
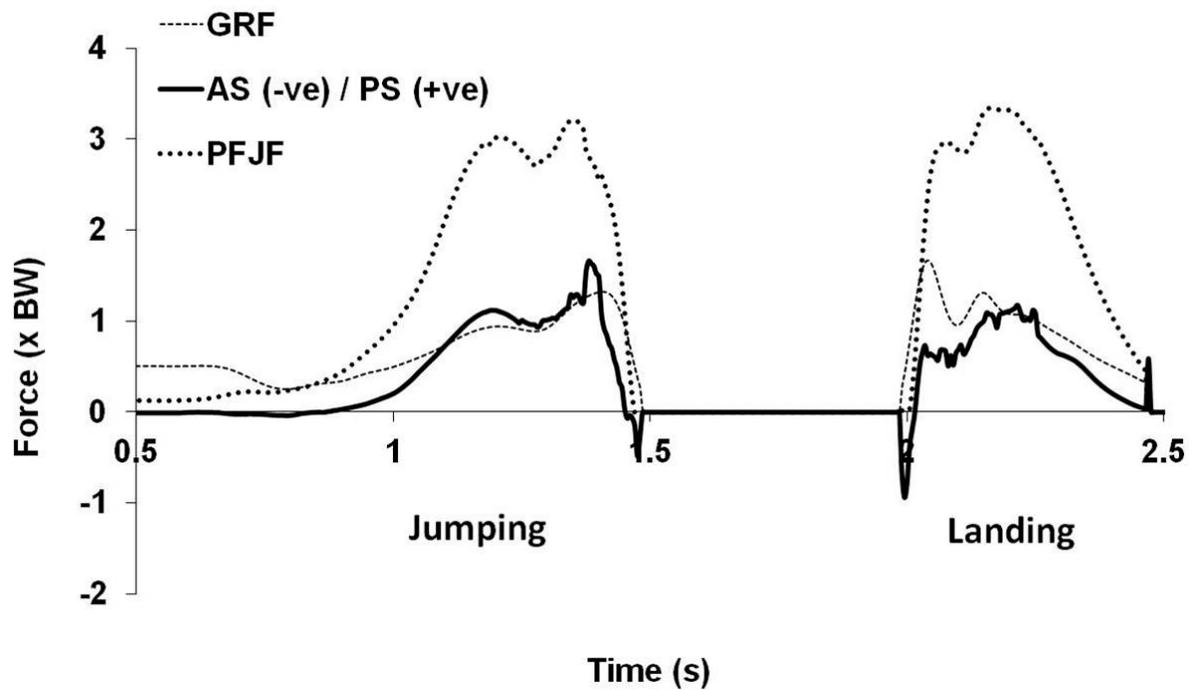
229

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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).
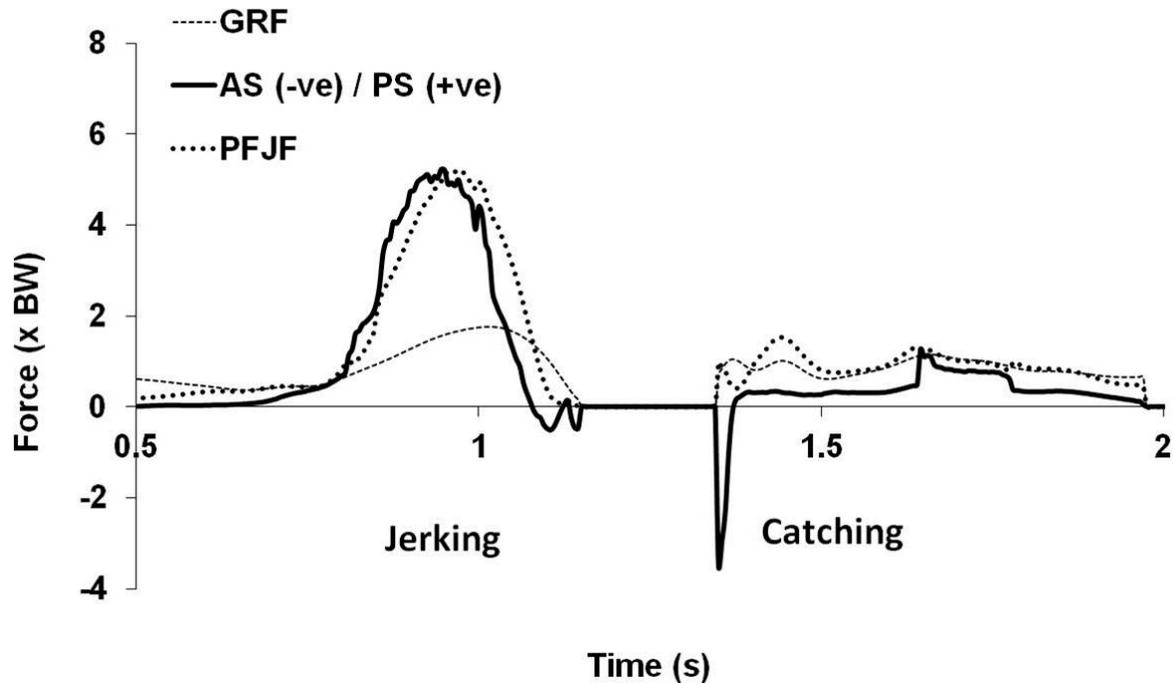
247

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 × BW, the ankle joint a loading of 8.9-10.0 × BW and the hip joint a loading

255   of 5.5-8.4 × BW. The peak PFJF was in the range of 2.4-4.6 × BW and the AS at the tibia

256   were in the range 0.3-1.4 × BW whereas the PS was of the order of 1.0-3.1 × 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 × 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**

316

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