A longitudinal study of impact and early stance loads during gait following arthroscopic partial meniscectomy

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Abstract

People following arthroscopic partial medial meniscectomy (APM) are at increased risk of developing knee osteoarthritis. High impact loading and peak loading early in the stance phase of gait may play a role in the pathogenesis of knee osteoarthritis. This was a secondary analysis of longitudinal data to investigate indices loading at baseline in an APM group (3 months postsurgery) and a healthy control group, and again 2 years later (follow-up). At baseline, 82 participants with medial APM and 38 healthy controls were assessed, with 66 and 23 re-assessed at follow-up respectively. Outcome measures included: i) heel strike transient (HST) presence and magnitude, ii) maximum loading rate, iii) peak vertical force (F_z) during early stance. At baseline, maximum loading rate was lower in the operated leg (APM) and non-operated leg (non-APM leg) compared to controls ($p \leq 0.03$) and peak F_z was lower in the APM leg compared to non-APM leg ($p \leq 0.01$). Over 2 years, peak F_z increased in the APM leg compared to the non-APM leg and controls ($p \leq 0.01$). Following recent APM, people may adapt their gait to protect the operated knee from excessive loads, as evidenced by a lower maximum loading rate in the APM leg compared to controls, and a reduced peak F_z in the APM leg compared to the non-APM leg. No differences at follow-up may suggest an eventual return to more typical gait. However, the increase in peak F_z in the APM leg may be of concern for long-term joint health given the compromised function of the meniscus.

Keywords: impact loading, medial meniscectomy

Introduction

Individuals following meniscectomy are at higher risk of developing early-onset knee osteoarthritis (Lohmander et al., 2007). This may be partly due to higher knee loads given that knee osteoarthritis is somewhat considered a pathological response of joint tissues to abnormal biomechanical stress (Englund, 2010).

Researchers have speculated that high impact repetitive loading may be involved in the pathogenesis of osteoarthritis (Felson and Zhang, 1998). Evidence from several animal studies *in vivo* support an association between repetitive high impact loads and articular cartilage damage

(Dekel and Weissman, 1978, Radin et al., 1978, Radin et al., 1973, Simon et al., 1972). Despite a paucity of studies investigating impact loads in humans with knee pathology, cross-sectional studies have found higher impact loading in people with knee osteoarthritis (Mundermann et al., 2005) and knee pain (Radin et al., 1991) compared with controls, albeit inconsistently (Henriksen et al., 2006).

Passive structures and active structures including the menisci and quadriceps act to dissipate and attenuate impact loads (Voloshin and Wosk, 1983, Jefferson et al., 1990). Given that meniscectomy impairs the ability to attenuate shock (Voloshin and Wosk, 1983), and quadriceps strength is reduced up to 6 months following APM (Glatthorn et al., 2010, Hall et al., 2013, Sturnieks et al., 2008a), these patients may experience higher impacts loads during gait. Determining if higher impact loads are present in people following APM, and how they may alter over time, can help direct and inform future research efforts aimed at reducing the risk of knee osteoarthritis in these individuals.

Impact loading is typically characterized in the vertical ground reaction force by a heel strike transient (Hunt et al., 2010, Mikesky et al., 2000, Simon et al., 1981, Smeathers, 1989, Verdini et al., 2006), maximum rate of loading (Hunt et al., 2010, Mikesky et al., 2000, Mundermann et al., 2005) and, while not impact-related, by the magnitude of the first peak vertical ground reaction force occurring early in stance has also been reported in this context (Hunt et al., 2010, Robbins et al., 2001). The first peak vertical ground reaction force reflects the maximum upward acceleration of the body’s center of mass, which may in turn reflect control of early, post-impact joint loading. The purpose of this exploratory study were to test the following hypotheses: 1) measures of impact loading and peak loading in the stance phase of gait are higher in the APM leg (operated) of people who have undergone a medial APM 3 months previously than healthy controls and the non-APM leg (contralateral leg) at baseline and 2 years later, and 2) longitudinal change in measures of impact loading and peak loading early in stance will be greater in the APM leg compared to controls and the non-APM leg.

Methods

Participants

This study reports on a secondary analysis of data from a prospective study described previously (Hall et al., 2013). Individuals aged 30-50 years with an isolated medial APM performed 3 months previously were recruited. Exclusion criteria included: evidence of lateral meniscal resection; greater than one third of medial meniscus resected; >2 tibiofemoral cartilage lesions; a single tibiofemoral cartilage lesion > approximately 10mm in diameter or exceeding half of cartilage thickness; previous knee or lower limb surgery (other than current APM); history of knee pain (other than that leading to APM); clinical or structural signs of osteoarthritis; post-operative complications; cardiac, circulatory or neuromuscular conditions; diabetes; stroke; multiple sclerosis and contraindication to magnetic resonance imaging. Healthy controls aged 30-50 years were recruited from the community. Exclusion criteria for controls included: current knee pain; knee injuries in the past year that either required medication reduced physical activity for at least one week, or resulted in time-off from work; previous knee arthroscopy that demonstrated the presence of osteophytes or involved diagnosis or treatment on a knee problem; cardiac, circulatory or neuromuscular conditions; diabetes; stroke; multiple sclerosis. Ethics approval was provided by the University of Melbourne Human Research Ethics Committee and written informed consent was provided by participants.

Gait Analysis

Gait analysis was performed using a custom seven-segment lower limb direct kinematics and inverse dynamics model (Besier et al., 2003). Kinematic (120Hz) and kinetic data (1080Hz) were collected using an 8-camera motion analysis system (Vicon, Oxford, UK) in synchrony with three force plates (AMTI Inc., Watertown, MA, USA). Participants performed five walking trials barefoot at a self-selected natural speed. Kinematic variables assessed at initial contact included: knee flexion angle, knee angular velocity, vertical ankle velocity (Hunt et al., 2010), and sagittal tibia inclination (with respect to vertical). Initial contact was defined as the moment where the vertical ground reaction (F_z) force reached above 20N.

The dependent variables were i) heel strike transient (HST); ii) max loading rate and; iii) the first peak F_z early in stance. Using techniques previously described (Hunt et al., 2010), a HST was determined to have occurred if during the upper 50% of the F_z slope, the F_z magnitude exhibited a transient peak (zero crossing in the first derivative), of a magnitude such that it subsequently decreased (trough) by >0.5% of the F_z first peak magnitude. The absolute magnitude of the HST was recorded. A participant was considered to display a HST if a HST was observed in at least 75% of trials (Hunt et al., 2010). Max loading rate was defined as the maximum rate of change of F_z, determined by the first order derivative of force versus time during weight acceptance. The first peak magnitude of F_z that occurred early in stance was also determined. The percentage of the stance phase at which the load-related variables occurred during stance was also recorded. All variables determined for each trial and then averaged across trials, with the load-related variables being normalized to body weight (N). Changes in the scores were calculated by subtracting the baseline from the follow-up scores. The mean of the right and left leg of controls were used in statistical analyses for continuous variables, while a randomly selected leg of the controls was used for the categorical HST variable.

Statistical Analysis

Independent t-tests were used to determine between-group differences in walking speed. For continuous variables, a mixed linear model was used to evaluate differences between legs (i.e. APM leg and controls; non-APM leg and controls; APM leg and non-APM leg of APM participants) at baseline and follow-up with 'participant' as a random effect and 'leg' as a fixed effect. Chi-squared tests were used to determine differences in the presence of HST between legs at baseline and follow-up. Longitudinal changes over time within each leg were assessed using paired t-tests and Wilcoxon signed rank tests as appropriate. A mixed linear model was used to assess differences in change between legs with 'participant' as a random effect and 'leg' as a fixed effect. The model was then adjusted for baseline scores (to account for baseline potential to change). When data did not conform to the Gaussian distribution, data were log-transformed prior to analysis. Stata 13.1 (Statacorp, College Station, TX, USA) was used for analyses and significance was set at $p<0.05$. With 82 APM participants and 38 controls at baseline, post-hoc analyses confirmed that we were powered to detect small to medium effect sizes (ES) for peak F_z (ES 0.30), maximum loading rate (ES 0.43) and HST magnitude (ES 0.37). Post-hoc analyses on the 2-yr change scores confirmed that the study was powered to detect small to large ES for peak F_z (0.46), maximum loading rate (ES 0.34), and HST magnitude (ES 0.72) between the APM leg and controls.

Results

Participants with an isolated medial APM (n=82) were assessed at baseline (88% male; age 41±5yrs; body mass index (BMI) 27.3±4.0kg/m^2) and 66 (79%) returned two years later for follow-up (86% male; age 41±6yrs; BMI 27.3±4.6kg/m^2). Thirty-eight healthy controls were assessed at baseline (84% male; age 41±7yrs; BMI 25.0±3.4kg/m^2) and 23 (61%) returned at follow-up (87% male; age 42±7yrs; BMI 25.5±3.4kg/m^2). The APM group had a significantly higher BMI (Hall et al., 2013) and walked significantly slower than the controls at baseline (p=0.015) (Table 1). However, walking speeds did not differ at follow-up. (Table 2)

Baseline

Vertical ankle velocity was lower in the APM leg compared to controls (p=0.03) and the non-APM leg (p=0.01) (Table 1) and, knee angular velocity also lower compared to controls (p=0.01) and the non-APM leg (p<0.001). Knee flexion angle at initial contact was greater in the APM leg compared to the non-APM leg (p<0.01). Tibia inclination at initial contact was more vertical at initial contact in the APM leg compared to controls (p<0.01). Maximum loading rate was lower in the APM leg compared to controls (p=0.03) and the non-APM leg (p=0.01). The time of maximum loading rate was earlier during stance in the APM leg compared and the non-APM leg (p<0.01), and later during stance in the non-APM leg compared to controls (p<0.01). Peak F_z was lower in the APM leg compared to the non-APM leg (p<0.01), and was no different compared to controls (p=0.054). The time of peak F_z occurred later in stance for the APM leg and non-APM leg compared to controls (p<0.01and p=0.03, respectively).

Follow-up

Knee flexion angle at initial contact was greater in the APM leg compared to the non-APM leg ($p<0.01$). Knee angular velocity at initial contact was lower in the APM leg compared to the non-APM leg ($p<0.01$). Tibia inclination was less pronounced in the APM leg at initial contact compared to the non-APM leg ($p<0.01$).The time of max loading rate was earlier during stance in the APM leg compared to controls ($p<0.01$) and the non-APM leg ($p<0.01$). The occurrence of a HST was lower in the non-APM leg compared to the controls ($p=0.02$).

2-year change

Walking speed did not change over time in either the APM group ($p=0.30$) or controls ($p=0.25$). Peak F_z increased in the APM leg compared to controls ($p=0.01$, adjusted for baseline score) and the non-APM leg ($p=0.01$, adjusted for baseline score). Of the 66 APM participants who returned for follow-up assessment, 35 of these individuals presented with a HST at baseline and follow-up in the APM leg. For these 35 participants, the magnitude of the HST increased over time ($p=0.01$) and increased compared to controls ($p=0.02$). However, there were no significant differences when baseline HST magnitude was taken into account.

Discussion

Key findings of this study are that 1) at baseline, the maximum loading rate was lower in the APM leg and non-APM leg compared to controls, and peak F_z was lower in the APM leg compared to the non-APM leg 2) at follow-up no differences in load-related measures observed between APM leg and controls, APM leg and non-APM leg, and non-APM leg and controls, and 3) the peak F_z significantly increased in the APM leg over 2 years compared to controls and the non-APM leg. These findings provide insight into parameters related to loading speculated to play a role in the etiology of osteoarthritis in people who have had APM and are at risk of developing knee osteoarthritis.

Contrary to our hypothesis, we observed a lower maximum loading rate in the APM and non-APM leg compared to controls at baseline. These results are inconsistent with previous research (Sturnieks et al., 2008a), that found no difference in maximum loading rate between the APM leg and controls in patients 3-months following surgery. Our findings are likely explained by the slower self-selected walking pace of the APM group compared to controls at baseline. Vertical ankle velocity and angular knee velocity at initial contact were lower in the APM leg compared to control and the non-APM leg. Although these differences in joint velocities may in part explain the lower maximum loading rate in the APM leg compared to controls, these kinematic differences between the APM and non-APM leg are not reflected in differences in maximum loading rate between the legs. Interestingly, the time at which the maximum loading rate occurred was earlier in the APM leg compared to the non-APM leg, and later in the non-APM leg compared to controls. However, these differences in timing are within 1% of stance time, and the clinical relevance of this minimal divergence in timing is unclear.

Normalized peak F_z that occurred in early stance was higher in the non-APM leg compared to the APM leg. Comparable to observations in first peak of the knee moments (Sturnieks et al., 2008b, Hall et al., 2013), the peak F_z, these results may suggest that people following a recent APM adopt a compensatory gait strategy that potentially attenuates loads in the operated leg. It is also interesting to note that there is an increased risk of developing osteoarthritis in the healthy, contralateral knee of people undergoing unilateral isolated meniscectomy (Englund and Lohmander, 2004).

Normalized peak F_z increased significantly in the APM leg over time compared to controls, such that there were no differences in peak F_z at follow-up, nor between the APM and non-APM legs. No differences at follow-up may reflect a return to typical gait. However, the increase in peak F_z in the APM leg is potentially concerning considering the compromised function of the medial meniscus in these patients and the repetitive nature of walking. Evidence from animal models suggests that high repetitive impact loads may lead to cartilage degeneration through subchondral microcracks (Burr and Radin, 2003). Studies using magnetic resonance imaging have reported abnormal subchondral and cartilage morphology in people following meniscectomy (Cicuttini et al., 2002, Mills et al., 2008, Wang et al., 2012, Williams et al., 2007). Recent evidence highlights the role of the posterior aspect of the medial meniscus in distributing load during the early phase of stance (Wang et al., 2014). It remains unknown however, whether the increases in peak F_z observed in this study exceed the physiological limits of the impaired joint structure and contribute to the development or progression of osteoarthritis that is often observed following meniscectomy.

Reduced quadriceps function is reportedly associated with higher rates of loading, greater presence of HSTs and HST magnitude (Jefferson et al., 1990, Mikesky et al., 2000). Interestingly, although we previously reported weaker quadriceps strength in the APM leg compared to controls (Hall et al., 2013), we observed a lower maximum loading rate in the APM leg compared to controls and no difference in HST presence or magnitude. Previous research found no difference in maximum loading rate between APM patients with normal and weak maximal quadriceps strength (Sturnieks et al., 2008a). Further evidence from people with knee osteoarthritis suggests that maximal quadriceps strength has minimal influence on measures of impact loading when walking speed is accounted for (Hunt et al., 2010). Collectively, the association between maximal knee muscle strength and maximum loading rate is questionable. It is conceivable that other aspects of quadriceps function are more related to impact loads during gait than maximal muscle strength.

There are several limitations of this study. Firstly, the differences in load-related variables may have existed prior to APM surgery and perhaps even contributed to the injury leading to the surgery. Therefore we cannot conclude that our findings are a result of APM surgery. Secondly, our lack of information about the cause of meniscal injury, and degree and location of meniscal resection for each APM individual is a limitation. The degree and location of meniscal resection reportedly alters the distribution of load across the tibial articular cartilage (Atmaca et al., 2013). Another limitation is that no adjustments were made for multiple comparisons and therefore caution must be used when interpreting our findings. We opted not to apply a statistical correction given the already increased type II error rate resulting from our reduced sample size at follow-up and exploratory nature of this study. Therefore, given the absence of statistical correction, caution must be used when interpreting our results. A further limitation is participant attrition from baseline to follow-up. Nonetheless, there were no significant differences for any of the load-related measures between those who did not return for follow-up assessment and participants who completed both assessments. Results of this study should be interpreted considering the absence of footwear and the use of self-selected walking speed as both are known to influence measures of impact loads (Hunt et al., 2010, Simon et al., 1981). Participants were instructed to walk at self-selected speeds and our statistical analyses were preformed on data unadjusted for walking speed to improve the external validity of our findings. However, results, adjusted for walking speed are available in the online supplementary material.

In conclusion, our study adequately powered to detect small to medium effect sizes, provides evidence that peak F_z increases over 2 years from 3-months following APM. Despite our statistically significant findings, the cross-sectional and longitudinal differences in peak F_z magnitude appeared low. However, walking is a repetitive daily activity and these low, yet significantly different magnitudes may be important. Further research is needed to understand the clinical significance and implications of increased peak F_z during gait on joint health, as therapeutic interventions may be designed if appropriate.

Conflict of Interest Statement

None of the authors have any conflict of interests' to declare.

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Table 1 Mean (95% CI) for the APM group and control group at baseline and follow-up

	Baseline			Follow-up		
	APM group (n = 82)			APM group (n = 66)		
	APM Leg	Non-APM Leg	Control (n = 38)	APM Leg	Non-APM Leg	Control (n = 23)
Walking velocity (m/s)	1.37 (1.34, 1.41)[a]	1.37 (1.34, 1.41)[a]	1.45 (1.40, 1.50)	1.39 (1.35, 1.43)	1.39 (1.35, 1.43)	1.43 (1.35, 1.50)
Vertical ankle velocity at initial contact (m/s)	-0.35 (-0.36, -0.34)[a, b]	-0.36 (-0.37, -0.35)	-0.37 (-0.39, -0.36)	-0.34 (-0.36, -0.33)	-0.35 (-0.36, -0.33)	-0.33 (-3.69, -2.90)
Knee flexion angle at initial contact (°)	3.62 (2.80, 4.45)[b]	2.00 (1.12, 2.88)	2.71 (1.54, 3.87)	4.27 (3.38, 5.17)[b]	2.90 (2.03, 3.77)	3.63 (2.56, 4.70)
Knee flexion angle velocity at initial contact (°/s)	157.6 (144.1, 171.1) [a, b]	187.9 (175.5, 200.4)	187.6 (170.0, 205.3)	181.0 (169.6, 192.4)[b]	195.8 (182.2, 209.4)	192.4 (174.0, 210.8)
Tibia inclination at initial contact (°)	-20.7 (-21.2, -20.2)[b]	-21.6 (-22.0, -21.1)	-21.1 (1.88)	-19.7 (-20.3, -19.2)[b]	-20.5 (-20.7, -21.0)	-20.5 (-21.0, -19.9)
Peak vertical force (BW)	1.18 (1.15, 1.20)[b]	1.21 (1.18, 1.23)	1.21 (1.18, 1.25)	1.21 (1.18, 1.23)	1.21 (1.18, 1.24)	1.18 (1.13, 1.23)
Time at peak vertical force magnitude (% stance)	22.0 (21.4, 22.6)[a]	21.6 (21.1, 22.1)[a]	20.5 (19.7, 21.4)	21.9 (21.3, 22.5)	21.9 (21.3, 22.5)	20.9 (19.2, 22.6)
Max loading rate (BW/s)	95.6 (88.2, 102.9)[a]	92.4 (85.3, 99.6)[a]	111.5 (97.9, 125.5)	101.5 (89.7, 113.3)	94.3 (85.3, 103.3)	106.0 (82.13, 129.9)
Time at max loading rate (% stance)	1.3 (1.3, 1.4)[b]	1.6 (1.5,1.6)[a]	1.5 (1.4, 1.6)	1.4 (1.2, 1.5)[a, b]	1.58 (1.52, 1.65)	1.60 (1.37, 1.82)
Occurrence of heel strike transient* n (%)	50 (61%)	46 (56%)	26 (68%)	41 (62%)	32 (48%)[a]	16 (70%)
Heel strike transient (BW)	0.73 (0.70, 0.77)	0.74 (0.71, 0.77)	0.78 (0.72, 0.84)	0.77 (0.73, 0.82)	0.78 (0.74, 0.82)	0.77 (0.66, 0.89)
Time at heel strike transient (% stance)	3.4 (2.9, 4.0)	3.0 (2.6, 3.5)	2.8 (2.3, 3.2)	3.8 (3.0, 4.5)	3.2 (2.7, 3.8)	3.5 (2.2, 4.8)

*The occurrence of a HST was recorded if present in ≥ 75% of the five trials (Hunt et al., 2010)

[a]$p<0.05$ compared to controls
[b]$p<0.05$ compared non-APM leg

Table 2. Mean (95% CI) change within legs, mean (95% CI) difference in change between legs unadjusted and adjusted for baseline scores

	Changes within leg Follow-up minus Baseline			Unadjusted changes between legs Follow-up minus Baseline			Adjusted changes between legs Follow-up minus Baseline		
	APM leg	Non-APM leg	Control	APM leg minus Control	APM leg minus Non-APM leg	Non-APM leg minus Control	APM leg minus Control	APM leg minus Non-APM leg	Non-APM leg minus Control
Vertical ankle velocity at initial contact (m/s)	0.00 (-0.01, 0.01)	**0.02 (0.00, 0.03)***	0.04 (0.00 to 0.09)	-0.04 (-0.07, -0.01)	-0.01 (-0.02, 0.00)	-0.03 (-0.06, 0.00)	-0.02 (-0.05, 0.00)	0.00 (-0.01, 0.00)	-0.02 (-0.05, 0.01)
Knee flexion angle at initial contact (°)	0.82 (-0.06, 1.70)	**1.15 (0.37, 1.94)***	0.48 (-0.35, 1.31)	0.34 (-1.17, 1.84)	-0.30 (-1.06, 0.45)	0.64 (-0.86, -2.15)	0.57 (-0.76, 1.91)	0.60 (-0.05, 1.24)	-0.02 (-1.37, 1.33)
Knee flexion velocity at initial contact (°/s)	**27.0 (18.0, 36.0)***	**0.11 (0.02, 0.20)***	14.2 (-1.85, 30.35)	12.7 (-4.44, 29.83)	**15.8 (3.92, 27.6)***	-3.06 (-20.3, 14.1)	3.98 (-10.9, 18.9)	4.67 (-5.09, 14.2)	-0.69 (-15.5, 14.1)
Tibia inclination at initial contact (°)	**1.09 (0.59, 1.59)***	**1.29 (0.84, 1.73)***	**0.74 (0.22, 1.27)***	0.37 (-0.49, 1.22)	-0.18 (-0.67, 0.32)	0.54 (-0.32, 1.40)	0.56 (-0.20, 1.33)	0.26 (-0.17, 0.69)	0.30 (-0.47, 1.07)
Peak F_z (BW)	**0.03 (0.01, 0.05)***	0.00 (-0.02, 0.02)	**-0.04 (-0.08, 0.00)***	**0.07 (0.03, 0.11)***	**0.03 (0.01, 0.04)***	0.04 (0.00, 0.08)	**0.05 (0.01, 0.09)***	**0.01 (0.00, 0.03)***	0.04 (0.00, 0.07)
Time at F_z magnitude (% stance)	-0.16 (-0.64, 0.31)	0.37 (-0.01, 0.74)	**0.83 (0.09, 1.58)***	-0.99 (-1.84, -0.15)	-0.53 (-0.94, 0.12)	-0.46 (-1.31, 0.39)	-0.60 (-1.44, 0.25)	-0.44 (-0.83, -0.04)	-0.16 (-1.00, 0.68)
Max loading rate (BW/s)	4.15 (-6.15, 14.46)	0.85 (-5.78, 7.48)	-9.40 (-23.10, 4.29)	13.56 (-2.64, 29.76)	3.43 (-3.97, 10.84)	10.13 (-6.12, 26.37)	9.76 (-6.37, 25.88)	4.22 (-3.07, 11.51)	5.54 (-10.75, 21.82)
Time at max loading rate (% stance)	0.04 (-0.06, 0.14)	0.04 (-0.01, 0.09)	0.10 (-0.13 , 0.33)	-0.06 (-0.23, 0.11)	0.00 (-0.10, 0.11)	-0.07 (-0.24, 0.10)	0.00 (-0.17, 0.17)	0.09 (-0.02, 0.20)	-0.09 (-0.26, 0.08)
Heel strike transient (BW)	**0.05 (0.01, 0.09)***	0.03 (-0.01, 0.06)	-0.03 (-0.10, 0.04)	**0.09 (0.02, 0.16)***	0.02 (-0.01, 0.06)	0.06 (-0.01, 0.14)	0.06 (-0.01, 0.13)	0.02 (-0.01, 0.05)	0.04 (-0.03, 0.11)
Time at heel strike transient (% stance)	**0.69 (0.13, 1.25)***	0.38 (-0.35 to 1.11)	0.24 (-0.37, 0.84)	0.44 (-0.64, 1.54)	0.16 (-0.40, 0.72)	0.29 (-0.82, 1.40)	0.70 (-0.41, 1.82)	0.14 (-0.36, 0.65)	0.56 (-0.57, 1.69)

F_z: peak vertical force; BW: body weight; ***bold values significance at p<0.05**

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A longitudinal study of impact and early stance loads during gait following arthroscopic partial meniscectomy

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